WEAK AT THE KNEES:

WEARABLE HAPTIC FEEDBACK DEVICE FOR REAL-TIME GAIT RETRAINING FOR SLOWING KNEE OSTEOARTHRITIS

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A thesis submitted in fulfilment of the requirements for the degree of Doctor of Philosophy,
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Abstract

Real-time gait retraining has been shown to be a promising non-invasive intervention for patients with early stage medial compartmental knee osteoarthritis. In order for real-time gait retraining to be effective beyond a clinical and laboratory setting, the technology needs to be effective but also portable.

This thesis presents a wearable haptic feedback ankle bracelet, which seeks to increase portability of real-time gait retraining systems. A gait retraining scheme is presented, involving the foot progression angle and step width, two gait parameters that are easily modifiable but also influential in reducing peak knee adduction moments (KAM), a surrogate measure for medial tibiofemoral joint loads.

Haptic feedback constitutes the first of the two main sections of this work, and is made up of two experiments. Two different haptic modalities were combined into a single device as a means of increasing portability of existing feedback systems while maintaining high user perceptual accuracy. The first experiment investigated lateral skin stretch as a modality for presenting directional feedback cues to participants on the lower limb. Findings show that human perception of directional skin stretch cues are extremely high, given the right speed, displacement and location of the stimulus. The second experiment explored the use of a haptic illusion, tactile apparent movement, as a modality for directional feedback cues on the lower limb. High direction perception accuracy was achieved when the participant received the optimal stimulus duration and interstimulus onset intervals during both standing and walking. The difficulties of integrating lateral skin stretch into a portable and wearable device compared with using tactile apparent movement is discussed. The proposed ankle bracelet combines tactile apparent movement with the widely used modality of binary (on/off) vibrotactile feedback into a single device; thus being capable of eliciting two different directional haptic cues to the user.

The second section of the thesis addresses the effectiveness of the ankle bracelet in the area of gait retraining and also consists of two experiments. The first experiment implemented the ankle bracelet as part of a real-time gait retraining task for the gait parameters foot progression angle and step width. Results showed that participants were able to retrain both gait parameters in a single short training session using the ankle bracelet. The type of feedback given (vibrotactile or tactile apparent movement) for either foot progression or step width (vice

versa) in general, did not matter. The second experiment investigated the use of a data-driven approach in order to determine participant-specific foot progression angle and step width modifications to reduce the first and second KAM peaks. Results showed that these models were not accurate enough to predict first and second KAM peaks. Despite this undesirable result, it was found that some participant-specific gait modifications altered peak KAMs, by reducing the second peak but increasing the first.

This thesis showed that select lower limb gait parameters can be easily trained using the proposed haptic ankle bracelet, improving portability over currently feedback devices. Even though more research is required in developing more accurate participant-specific predictive models of KAM peaks; a scheme involving foot progression angle and step width remains promising as they were able to significantly alter peak KAMs. Real-time haptic gait retraining is an intervention, that has the potential to guide participants to change their walking patterns, which could effectively slow the progression of knee osteoarthritis.

Acknowledgements

> "You take the blue pill, the story ends. You wake up in your bed and believe whatever you want to believe. You take the red pill, you stay in Wonderland, and I show you how deep the rabbit hole goes."
>
> —Morpheus, to Neo[1]

I remember telling myself (and God) during undergrad, I'd never be so foolish as to pursue a PhD. Well, I am writing this from inside the rabbit hole; though I'm unsure of how deep this rabbit hole really is and where it ultimately leads, it definitely has been an exciting journey thus far.

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I laughed a ton, was weirded out occasionally, and was helped beyond measure frequently. I loved how we were able to do life together; from using the workshop for making our personal projects, having lunch in a sketchy bar in Tijuana and rushing back to San Diego before dusk, exploring the beauty of Yosemite, having pizza and beers, and playing a ridiculous amount of board games in the lab. I don't think I'd have ever thought research could be so enjoyable.

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Allan Veale, he deserves special mention because I don't think I would've been able to do a PhD if it were not for him. I don't think I've met anyone as hardworking, intelligent and humble as Allan; the fact he agreed to be my project partner for virtually every group project during undergrad is still beyond me. If I didn't understand something, I knew Allan would

probably have the answer. Allan's work ethic is enviable. He made sure he did everything to the best of his ability and then some. I look forward to future collaborations if we both remain in academia!

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[1] The Matrix. Dir. Andy Wachowski and Larry Wachowski. Warner Bros. Pictures, 1999. DVD.

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The undersigned hereby certify that:

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Chapter 1: Introduction

1.1 Purpose

Miniaturisation of electronics, extended battery life, and advances in computing power in recent decades has accelerated the potential of integrating technology with medical and rehabilitation interventions. One such field has been that of real-time gait retraining whereby sensors, computer algorithms, biomechanical models, and haptics, synergistically form part of a complete system for guiding individuals to a new gait. The intention of this is to reduce the knee adduction moment (KAM), a surrogate measure of the knee joint force, which has been linked to the progression of medial compartmental knee osteoarthritis (OA) [1-3]. Presently, wearable haptic feedback devices have been successfully implemented on the foot [4, 5], shank [4], forearm [6], head [7-10], torso [11], and trunk [4, 12, 13] to retrain gait parameters like foot progression angle (FPA) [4], tibia angle [4], and trunk sway [4, 7, 8, 10]. Despite current prototypes successfully retraining one's gait, their inherent design of needing to attach different devices to various body segments may prove to be too cumbersome for patients to integrate into their daily lives as part of any real-time gait retraining program. Therefore, some requirements, which have been identified in the field of wearable robotic orthoses that the author feels can be included in the area of haptic gait retraining are: effectiveness, comfort/acceptance, and portability [14].

The purpose and motivation of this work is to:

- Develop a portable real-time haptic device capable of giving multi-modal haptic feedback to alter foot progression angle and step width.
- Explore the effectiveness of this real-time feedback device on retraining foot progression width and step width, and whether or not peak KAMs can be reduced using this retraining scheme.

The haptic device proposed in this thesis could be used in future real-time gait retraining systems and act as a non-invasive intervention for early medial compartmental knee osteoarthritis.

1.2 Knee Osteoarthritis

This section gives an introduction to knee osteoarthritis, its impact, and the interventions which exist and are presently used to treat this disease.

1.2.1 What is Knee Osteoarthritis?

Knee osteoarthritis is a common joint degenerative disease whereby the articular cartilage of the femur, patella, or tibia breaks down and no longer performs its function of providing a frictionless bearing and distributing load to the underlying bone. For the purpose of this thesis, only tibiofemoral joint OA will be considered, which involves degeneration of the femoral or tibial cartilage. Repetitive mechanical loads experienced during walking gait are believed to be play an important role in the onset [15], progression and severity of the disease [2, 3]. The medial compartment of the tibiofemoral joint is known to be 64% more likely for cartilage lesions to progress compared with other regions of the knee [16]. As there is no cure for OA (articular cartilage has poor ability to regenerate and spontaneously heal), treatments of knee OA have focused on pain reduction, maintaining function, and slowing its progression. This approach is warranted as end-stage sufferers of knee OA require total joint replacement, which is a costly and invasive surgery.

1.2.2 Impact and Prevalence of Knee Osteoarthritis

In New Zealand, arthritis (including rheumatoid) is estimated to cost $3.2b, representing 1.7% of the country's total gross domestic product [17]. The self-reported prevalence of OA in New Zealand based on one survey was 7.7% [17]. In 2010, the public inpatient costs in New Zealand was estimated to be $99.9m, 42% of total hospital costs, and knee OA represented a 31% share, second to hip OA [17]. Studies have shown that healthcare costs of those with OA significantly exceed those without and is disproportionately higher for women [18]. Sayre et al. [19] found that knee OA was a significant predictor in whether or not one had a reduction in employment. A survey in Canada revealed that many indirect costs are also associated with OA incurred mainly from the time lost from employment and leisure by patients and their caregivers [20].

Aside from the economic effects of knee OA, sufferers also face reduced quality of life compared with healthy peers [21]. No other disease acts as a greater obstacle for functional activities like walking and climbing stairs than OA [22]. Patients with knee OA also report

having difficulties with common activities such as housekeeping and carrying physical loads [23]. Apart from the physical aspects on quality of life, the Short-Form Health Survey (SF-36), an instrument for assessing health-related quality of life [24], also evaluates areas such as mental, emotional, and vitality of the person. Using the SF-36, there is evidence that older adults with lower limb OA not only have poorer physical health, but lower mental health too [21].

The Framingham study indicates that 11% of women and 7% of men aged 63 to 75 years are affected by the symptoms of knee OA. However, that percentage is around three times higher when taking into account the number who show radiological evidence of knee OA [25]. A cross-sectional health examination survey in the US (NHANES III), estimated prevalence of radiographic knee OA in persons aged 60 years and over and concluded that symptomatic knee OA affects more than 10% of adults in the US of that age group [26].

1.2.3 Diagnosis

Traditionally, osteoarthritis was classified strictly using radiographs and described using the Kellgren and Lawrence (K/L) scale [27]. The K/L scale is divided into five grades:

(0) No evidence of osteoarthritis
(1) Doubtful whether disease exists
(2) Definite evidence of osteoarthritis but minimal severity
(3) Moderate disease with signs of osteophytes and joint space narrowing
(4) Severe evidence of osteoarthritis, which may include osteophytes, narrowed joint spacing, presence of cysts, and altered shape of the bone ends.

Relying on radiographic evidence, however, does not seem sufficient in some cases. Classically, OA would be identified if a joint had a disease of grade 2, having evidence of minimal osteophytes. One problem as mentioned by Spector et al. is that while it is possible to have evidence of osteophytes with no narrowed joint spacing (grade 2), the reverse case of having narrowed joint spacing without the presence of osteophytes was also possible, allowing for the opportunity of misclassification [28]. Radiographic evidence alone also does not take into consideration the functional impact and pain, which may not be consistent with radiographic criteria [22].

Symptomatic knee OA has thus been defined as showing both radiographic and clinical evidence of knee OA. Clinical evidence is evaluated through the use of surveys such as the Western Ontario McMaster Universities Arthritis Index (WOMAC) survey, which encompasses questions for pain, stiffness, and for function [29].

1.2.4 Interventions

A review conducted by the Osteoarthritis Research Society International (OARSI) of current therapies for the management of hip and knee OA and the level of evidence supporting the efficacy of each divided interventions into three main categories (not including combination therapies): pharmacological, non-pharmacological, and surgical [30].

Although pharmacological therapies have shown evidence to be effective in the management of pain caused by knee OA, many have negative gastrointestinal [31], cardiovascular [32], and diarrhoea inducing [33] side effects [30]. Kon et al. [34] suggested that caution should be exercised when using pharmacological agents in the management of knee OA and that it should be considered with other conservative measures.

Surgical interventions for early medial compartmental knee OA such as unicompartmental knee arthroplasty and high tibial osteotomy (HTO) can be performed [35]. In unicondylar knee arthroplasty, an implant is used to replace the contact surfaces of the affected part of the knee. Studies show that the 10 year survival rate for a unicondylar knee implant can range from 80.2 to 98% [36, 37]. In HTO, a wedge of bone is taken out from the tibia and held in place by a metal plate and screws in order to reduce the loading on the medial side of the knee joint. HTO reduces the KAM [38, 39], and shows evidence that it reduces pain and improves function [30, 40, 41]. However, surgical treatments are not without their risks, which include infection and the need for revision surgeries in the future [42-45].

Non-pharmacological treatments of early knee OA include lifestyle changes, exercising and physical therapy [34]. Conservative approaches should be the first treatments sought as they avoid the potential risks involved with pharmacological and surgical interventions. Kon et al. [34] surveyed non-surgical methods of managing early knee OA and found that exercising and physical therapies to be viable options. The area of conservative treatment which has shown promise in recent years has been that of gait retraining for reducing the KAM, a review of which can be found by Richards et al. [46]. It presents a non-invasive and low risk way of

potentially slowing the progression of early medial compartmental knee OA; although the long-term effects of gait retraining on cartilage degeneration are yet to be observed.

1.3 Gait Retraining

Gait retraining is described in this section, specifically its application in the area of slowing the progression of knee OA. This section also introduces the concept of the knee adduction moment, or KAM, its purpose in gait retraining, and the three aspects that encompass a complete gait retraining system.

1.3.1 What is Gait Retraining?

Gait retraining is the modification of gait parameters while walking or running in order to achieve a more optimal state with respect to some rehabilitation and sporting application. Other fields where this has been applied, apart from the application described in this thesis, have been in the treatment of patients following stroke [47], with cerebral palsy [48], or even runners reducing tibia stress [49]. A key element of gait retraining is selecting an appropriate kinematic or kinetic variable to modify. In terms of gait retraining for knee OA, the variable that has received most attention is the KAM.

1.3.2 The Knee Adduction Moment

The fundamental premise of altering walking gait to influence the progression of medial compartment knee OA is to reduce the load on the medial aspect of the joint, thus reducing the progression of articular cartilage degeneration. Unfortunately, the internal loading of the medial compartment of the knee cannot be measured directly and must be estimated using computational modelling of walking dynamics. However, the applied KAM during the stance phase of gait presents a simple, surrogate measure of the medial-lateral load distribution at the knee [1], which can easily be measured in a gait laboratory using a force plate and motion capture system. Although it might not be a direct measure of medial compartment loading, the KAM has been associated with the onset, progression and severity of knee OA [2, 3], making it a logical surrogate for use in gait retraining. The main strategy therefore in gait retraining for patients with knee OA has been to reduce the KAM during the stance phase of walking.

1.3.3 Real-Time Gait Retraining System

Having provided the context and purpose of gait retraining for patients with knee OA, it is important to discuss the elements which constitute a functional real-time gait retraining system. For gait retraining to be a useful intervention outside of clinical and laboratory settings, it has to be capable of measuring patient-specific kinematics and/or kinetics, and give feedback in real-time. There are three functional blocks within a real-time gait retraining scheme (Figure 1.1):

i. Defining optimal gait kinematics
ii. Sensing of patient kinematics
iii. Feedback to the patient, guiding them to alter certain gait parameters to a more optimal state

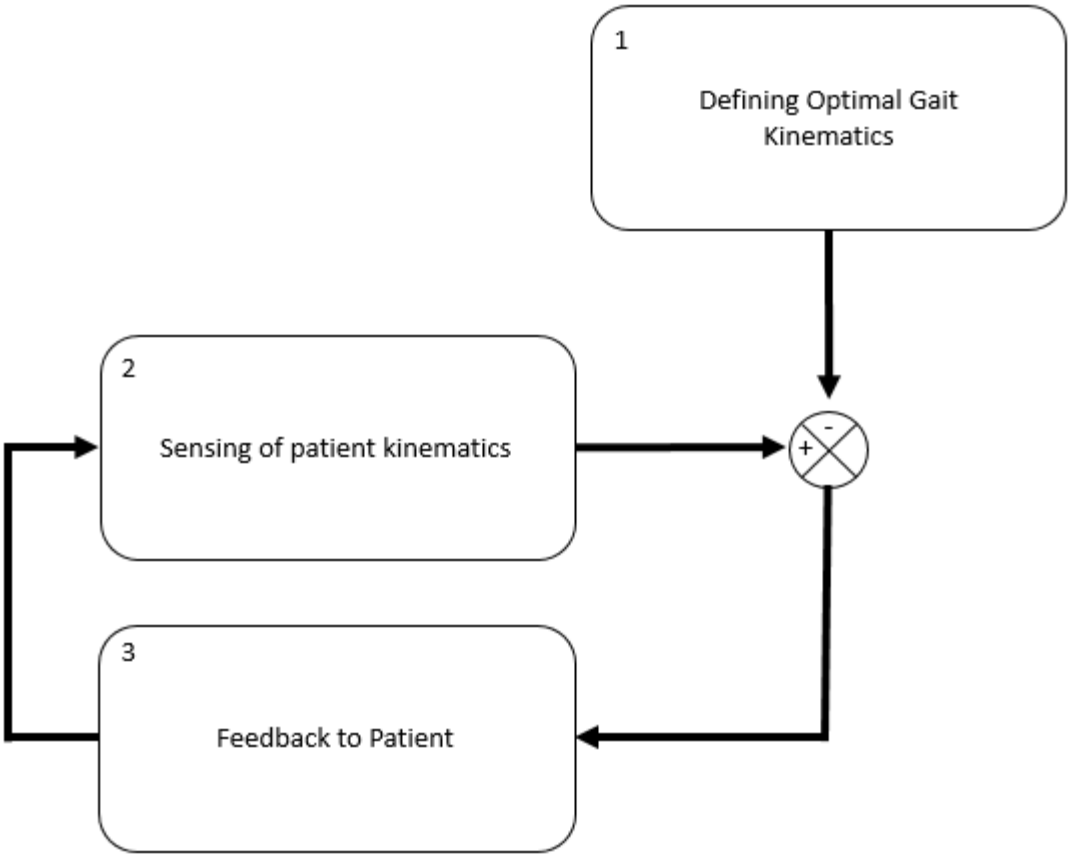

Figure 1.1: Block diagram showing the three aspects in a gait retraining scheme. Once an optimal gait is determined (1), the patient's walking gait is analysed in real-time to estimate his or her kinematics (2). These data are compared to the ideal model and used to generate a feedback signal, which is then provided to the patient (3), such that they can adjust their walking gait on the next step.

1.3.3.1 Defining Optimal Gait Kinematics

The first functional block of a gait retraining scheme involves defining the optimal gait parameter(s) to be adjusted. In this case the KAM during the stance phase of gait provides a useful, easy-to-measure 'surrogate' of tibiofemoral joint contact force. The main function of the gait retraining scheme is to define a new walking gait that minimises the KAM. The challenge is to identify modifiable gait parameters that influence the KAM, which are typically kinematic parameters, such as trunk angle, or foot progression angle [50]. Kinematic variables are highly variable within and across individuals and their influence on the KAM is also variable and non-linear. Hence, a patient-specific calibration or testing protocol is necessary to identify the relationship between the altered gait parameters and KAM. The desired gait parameters are then inputs, which are continuously compared with the real-time patient kinematics.

1.3.3.2 Sensing of Kinematics

Presently, the gold standard to obtain patient kinematics and ground reaction forces is with a motion capture system and a static force place. Although these tools for motion sensing have been used in real-time gait retraining exercises successfully, they are costly and confined to a clinical and laboratory setting, which inhibits implementing real-time gait retraining in people's daily lives. Hence, there has been an ever developing field of portable motion capture technologies encompassing both hardware and software which seeks to remove these limitations. Shoe insoles have been developed, capable of measuring foot force and pressure [51]. Instrumented shoes with embedded miniature force plates have also been designed to obtain similar information to that of instrumented treadmills like three axes force and moments [52]. Likewise, to capture the kinematics by using accelerometers, gyroscope, magnetometer hardware packages like inertial measurement units [53, 54] and sensor algorithms [55, 56], have been synergistically combined into wearable devices for motion capture.

1.3.3.3 Feedback to Patient

Many different methods of feedback for gait retraining exist, interacting with different sensory channels on the body; including visual [7, 8], auditory [57, 58], and haptic (touch) [4, 6]. Despite visual stimuli being a powerful form of feedback for retraining gait, previous implementations required the use of visual displays, which restricted the portability and use

[59, 60]. A review by Veale et al. stated that comfort and acceptance were important requirements for a wearable orthosis in order to avoid stigmatization [14]. Auditory feedback, although being able to be implemented discretely through the use of earphones; suffers from the same caveat as visual feedback in the sense that it requires the patient to focus their attention on the audio cues, which would normally be reserved for interacting with their surroundings. An increase in the cognitive load may also be a factor in using visual and auditory sensory feedback especially in elderly patients [61, 62]. Haptic feedback thus presents an attractive alternative in effectively providing gait retraining cues but also allowing for the potential for devices to be concealed under clothing during use without distracting visual and auditory sensory channels during daily walking.

1.4 Thesis Overview

This thesis proposes a new wearable haptic ankle bracelet which can be used to simultaneously retrain two gait parameters, namely foot progression angle and step width, and explores how well retraining those parameters reduce the peak KAM in a single participant. The work is organised as follows (also illustrated in Figure 1.1):

Chapter 2 provides the background literature surrounding haptic feedback, specifically the basic mechanisms of sensing mechanical stimuli via the human skin, different haptic modalities that have been explored and implemented, and gait retraining with a focus on kinematic gait parameters. I also highlight areas that are currently unknown in this field, paving the way for the research questions set out in the beginning of the introduction.

The concept of combining multiple haptic modalities into a single device is introduced in Chapter 3. Chapter 4 explores the perception of lateral skin stretch on the lower limb, detailing the accuracy rates of determining the direction of a stimulus at different displacement, speeds and locations. This section explores whether or not lateral skin stretch would be an effective feedback modality for providing direction feedback cues. Chapter 5 explores the use of a haptic illusion called tactile apparent movement (TAM) as a modality for feedback. The main parameters inter-stimulus and stimulus durations for optimal perception of movement, directional perception accuracy, and response times during standing and walking were investigated. Chapter 6 discusses the two modalities that were explored and the advantages and disadvantages of implementing them into a multi-modal haptic feedback device for the lower limb.

Chapter 7 is a study where tactile apparent movement was combined with binary vibration into a multi-modal feedback system and applied to a multi-parameter gait retraining task of foot progression angle and step width targeting walking on the treadmill. This study looked into whether or not an individual could retrain two parameters in a single training session (< 15 mins) and whether or not the gait parameter and modality (binary vibration or TAM) combination made a difference in the training time.

Chapter 8 is a study which used a data-driven approach to find new foot progression angle and step widths to reduce the KAM. An individual was chosen based on initial scatter matrices, which showed promising correlations between the gait parameters and the KAM peaks, and regression models predicting the first and second KAM peaks were constructed. Actual values of first and second KAM peaks at the new gait recommendations were measured and compared with the model predictions. The study shed light on whether the methodology was suitable to reduce peak KAMs.

Chapter 9 contains the concluding remarks, summarising the results with respect to the research questions initially posed. Future work in this field and in the areas touched on in this thesis are mentioned.

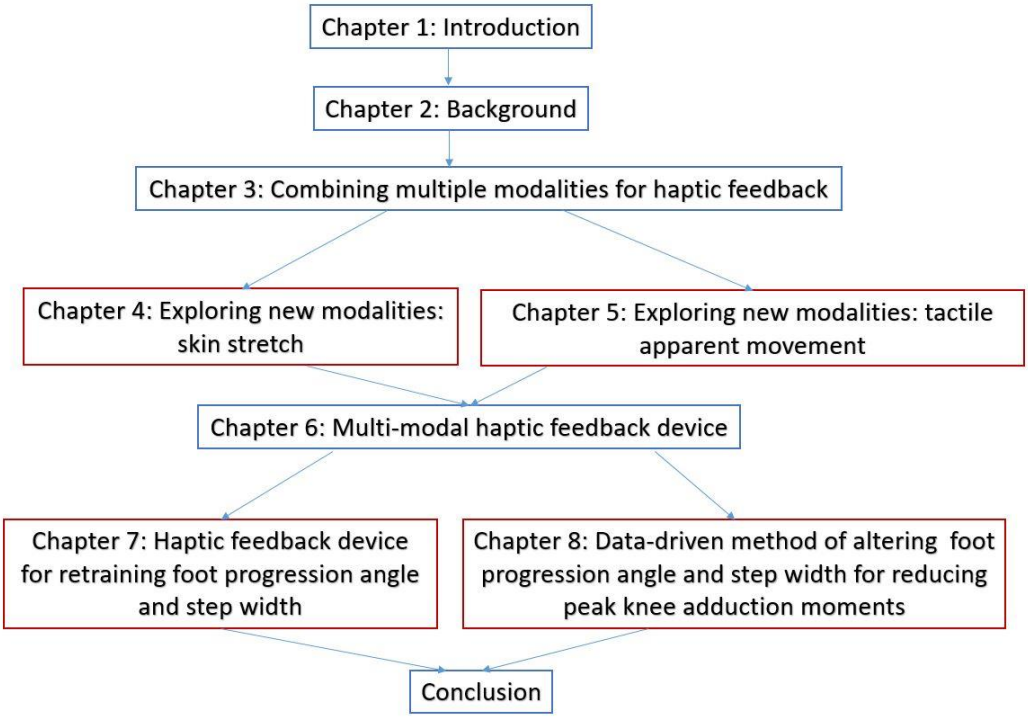

Figure 1.2. Flowchart showing the overview of the different chapters of the thesis. Red boxes illustrate chapters of the thesis which are the novel contributions to the thesis.

1.5 Thesis Contributions

The main contributions resulting from this thesis in the field of wearable haptic feedback for reducing the knee adduction moment includes:

- *Lower Extremity Lateral Skin Stretch Perception for Haptic Feedback.* This work, although wasn't used in the proposed haptic device, showed promising results as a new modality for providing directional feedback cues to the lower limb. High accuracy rates were achieved and optimal locations on the lower limb were highlighted. Although the author did not feel it was suitable for a wearable and portable device, it is definitely viable in its current state if it was implemented in a grounded stationary feedback system.
- *Tactile Apparent Movement as a Modality for Lower Limb Haptic Feedback.* A novel implementation of a haptic illusion was implemented on the lower limb with its optimal parameters identified during standing and walking scenarios. This modality was chosen as the one to combine with binary vibration into a multi-modal feedback device due to its ease of integration but also its high accuracy rates and low response times suitable for real-time gait retraining applications.
- *Wearable lower limb haptic feedback device for retraining Foot Progression Angle and Step Width.* This study was a novel implementation of using two different feedback modalities in a single device to retrain the two gait parameters, foot progression angle and step width. This showed that using the proposed haptic device, participants were able to intuitively walk with a new gait in a single training session under 15 mins.
- *Data-driven Method for Reducing Peak Knee Adduction Moments by Altering Foot Progression Angle and Step Width.* This work presents an initial attempt at determining foot progression angle and step width modifications that reduce the peak knee adduction moments. The approach used here was unable to accurately predict the knee adduction moments, which the author suspects is due to the walking methodology employed during the data collection stage. The gait recommendations had an adverse effect on the first KAM peak but reduced the second.

The author was responsible for the majority of the work in all four studies listed above. This includes the mechanical, electrical, software design, study design, data acquisition, data analysis, and paper write up. The co-author for chapter 7, Markus Haller, was responsible for assembling the ankle bracelet for that particular study.

Chapter 2: Background

This chapter gives an overview of the fundamental concepts and reviews the literature that surrounds the research areas of interest in this thesis. The two main fields which will be discussed here include haptic feedback and gait retraining, in the context of reducing the KAM during the stance phase of walking.

2.1 Haptic Feedback

Haptics is the science of the sense of touch which encompasses the two main areas of kinaesthesia, the ability of perceiving the position and motion of a body in space; and the cutaneous sense which is the act of sensing pressure, temperature and pain [63, 64]. Haptic feedback is the act of providing a cutaneous and or a kinaesthetic stimuli to the intended individual. Tactile feedback is a subset of haptics, and even more specifically cutaneous feedback but is only focused on the sensations of pressure via mechanoreceptors on the skin [63, 64]. A tactile feedback device or tactile display is used to provide the individual with some kind of pressure related mechanical stimulus. Although haptic feedback is a broad term referring to the definition above, in the field of gait retraining it is understood to be in the form of tactile feedback. To be consistent with the literature [6, 46, 65], haptic feedback will be the term used throughout this thesis.

2.1.1 Mechanoreceptors

The human skin is embedded with many sensors called mechanoreceptors. Different mechanoreceptors exist depending on whether the skin is glabrous (devoid of hair) or non-glabrous (hairy). Fast-adapting (FA) and slow-adapting (SA) are the two basic types of mechanoreceptors. FA receptors activate at the onset and cessation of a stimulus but do not fire during the presentation of a sustained stimulus. SA receptors activate at the onset of a stimulus and continue to do so until the stimulus is removed. Mechanoreceptors in the human skin are further categorised based on their receptive fields, namely type I and type II. Type I receptors have small receptive fields with clear boundaries and are typically located closer to the surface of the skin, whereas type II receptors have large receptive fields with poorly defined boundaries and are located deeper within the dermis of the skin [66, 67].

Extensive research has been performed on glabrous skin (e.g. palm, fingers) and four types of mechanoreceptors: FAI, FAII, SAI, and SAII, have been identified [68]. Different end organs are associated with the four types of mechanoreceptors. In glabrous skin, FAI units are connected to the Meissner's Corpuscles, FAII to the Pacinian corpuscles, SAI to the Merkel cell, and SAII to the Ruffini endings [69], illustrated in Figure 1.1. Substantially less is known about the receptors in hairy skin, although SAI, SAII, and Pacinian units all exist with the addition of hair units and field units [70]. Table 1.1 summarizes the mechanoreceptors along with their characteristics.

2.1.2 Modalities

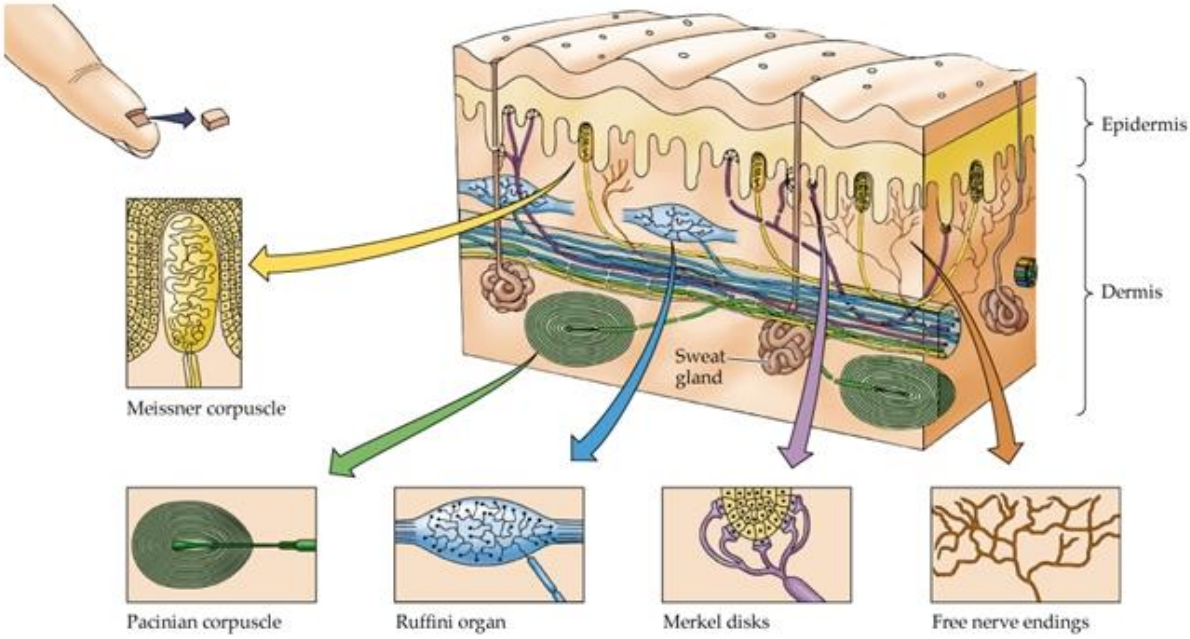

Figure 2.1. Mechanoreceptors found in glabrous (non-hairy) skin. Image from [http://grants.hhp.coe.uh.edu/clayne/6397/Unit4a.htm].

2.1.2.1 Vibrotactile

The most common haptic modality that has been used as a form of feedback to the human skin has been vibration. The Pacinian corpuscle can sense the lowest known deformation threshold out of all other receptors with peak sensitivity when the stimuli has a frequency of around 250 Hz [71]. The threshold at different frequencies have been previously investigated at numerous locations on the body [72-75], with the fingers showing the highest sensitivity of 0.07 µm at 200 Hz [76]. When the skin is normally displaced at a slow rate (< 0.15 mm/s in fingers) the SAI receptor types are activated instead of the Pacinian corpuscles and the threshold of displacement also rises [77].

Both temporal and spatial characteristics of vibratory stimuli are important in the perception of a haptic cue. Three temporal components which apply to vibrotactile stimuli are the stimulus duration, pulse repetition, and number of pulses [72]. Identifying a tactile stimulus improves as the stimulus increases in duration from 80 to 320 ms [78]. Tactile pulses in the range of 50 to 200 ms are most preferred as longer durations are perceived as irritating [79] and can lead to desensitisation [80]. Temporal parameters of a vibrotactile stimulus are important for perceiving the stimulus intensity and differentiating between different haptic cues. However, spatial parameters are important for localising and encoding directional information. The ability to localise the stimulation from a tactile array is highly dependent on the intertactor spacing [72]. An important aspect of recognising individual vibratory stimuli arising from a tactile array is the intertactor spacing must be greater than the two-point discrimination threshold for vibrations. Despite a lack of data in the literature, Eskildsen et al. [81] discovered that the two-point threshold on the back for vibrotactile stimuli is around 11 mm, significantly smaller than that of the static two-point threshold which is around 20-40 mm [82].

Table 2.1. Summary of mechanoreceptors along with characteristics as suggested by Vallbo et al., Verrillo, and, Jones and Sarter [70-72].

Receptor	Receptor type	Sensory Correlation	Receptive field size	Frequency range	Threshold skin deformation (median)
Pacinian Corpuscle	FAII	Vibration	Large	40-400Hz (most sensitive ~250Hz)	9.2µm
Meissner's Corpuscles	FAI	Stroking and fluttering	Small	10-200Hz	13.8µm
Hair follicle	FAI	Hair displacement	Large	?	?
Ruffini ending	SAII	Skin stretch	Large	~7Hz	331µm
Merkel's disks	SAI	Pressure	Small	0.4-100Hz (most at ~7Hz)	58.5µm
Field Receptors (not shown)	FA	Skin stretch and joint movement	Large	?	?

Various illusory phenomena exist, which have been previously explored and have the ability to convey directional information. The most common haptic illusion is that of saltation, whereby a few successive pulses stimulated at discrete sites create the illusion of the stimulus 'hopping' from one site to another [83-89], as illustrated in Figure 1.1. The separation, stimulus

duration, interstimulus interval, locus, and intensity are all influential parameters in whether or not saltation is convincingly perceived.

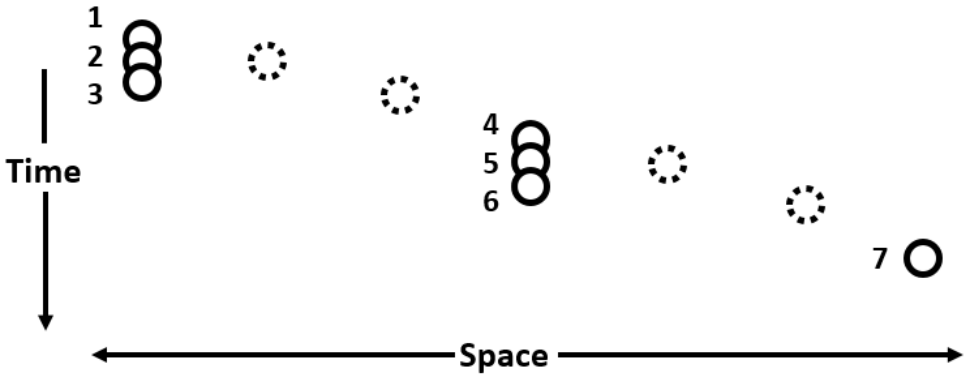

Figure 2.2. Visual representation of saltation where three tactors pulsed in the pattern shown above creates the illusion of a continuous stimulus running from the first tactor to the third. Illusion is depicted by the dashed circles. Figure adapted from [91].

A much less commonly implemented haptic illusion, tactile apparent movement (or motion) (TAM), occurs when tactors are sequentially activated along a discrete array creating the illusion of a continuous stroking sensation [90]. Like saltation, both temporal and spatial parameters of the stimuli are vital to the perception of this illusion. The two temporal parameters which have been shown to be most important for apparent movement is the interstimulus onset interval (ISOI) and the stimulus duration (SD), illustrated in Figure 1.1 [90-94]. Spatially, increasing the number of stimulators improves the apparent movement [94], but the interstimulus distance [91] and the site of stimulation does not seem to have a profound influence [89].

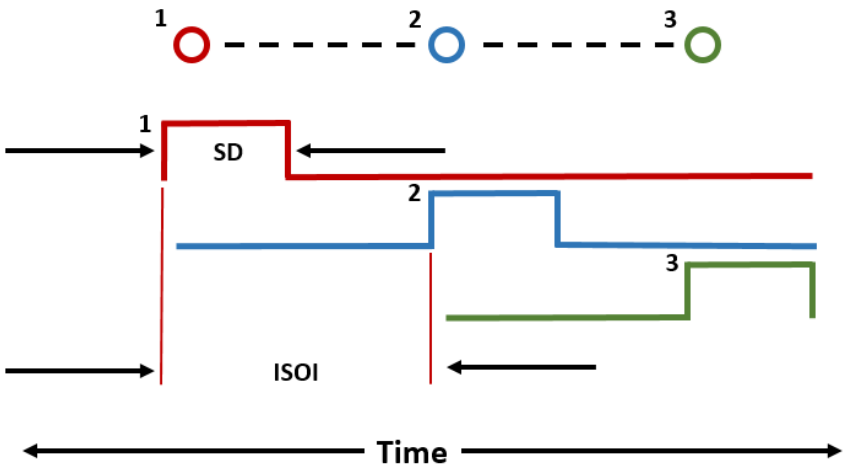

Figure 2.3. Visual representation of tactile apparent movement with three tactors. Stimulus duration (SD) is depicted by the square waves and interstimulus onset interval (ISOI) is the time interval separating the beginning of each tactor.

2.1.2.2 Skin Stretch

A tangential or lateral skin stretch is dominated by the SAII mechanoreceptor type [95], and has more recently been explored as a haptic feedback modality. SAII units tend to be clustered in human nerves [96], more specifically the peripheral nerves [97, 98] and exist in smaller proportions compared with other mechanoreceptors [99, 100].

Reasons as to why skin stretch is an appealing modality is that tangential displacements are more perceivable compared with normal displacements in hairy skin [101]. Skin stretch is directionally sensitive [102, 103] and previous literature has shown it has a lower direction detection threshold than for a spatiotemporal stimulus like stroking [104]. In order for directional cues to be encoded using vibrotactile feedback, an array of actuators is generally required. However, because lateral skin stretch is inherently directional, a single actuator can theoretically perform the desired feedback.

Despite skin stretch having advantages in providing directional haptic feedback, there are also practical considerations when this modality is to be implemented as a wearable device. The perception of skin stretch is highly dependent on multiple factors including the normal force applied during stretch [105], the rate of stretch, and the skin stiffness [106]. Previous experiments exploring the displacement and speed relationship during lateral skin stretch have glued the stimulator to the skin to prevent slip, which may alter its suitability when applied to a wearable haptic device [101, 107]. The majority of prior literature studied skin stretch for the upper limb [108-119] with a few exceptions like Wasling et al. [120] who was interested in somatosensory cortex activity during skin stretch. Little is known about the application of lateral skin stretch as a haptic device for the lower extremity.

2.2 Gait Retraining for Reducing the Knee Adduction Moment

The main strategy in gait retraining which has been used to slow the progression of medial compartmental knee OA has been reducing the KAM. The KAM is defined as the moment occurring at the knee joint centre (KJC) resulting from the ground reaction force (GRF) at the foot's centre of pressure (COP) in the frontal plane of the tibia (seen in Figure 1.1). It can be calculated by taking the cross product of *r*, the position vector from the KJC to COP, and the *GRF* vector:

$$KAM = r \times GRF$$

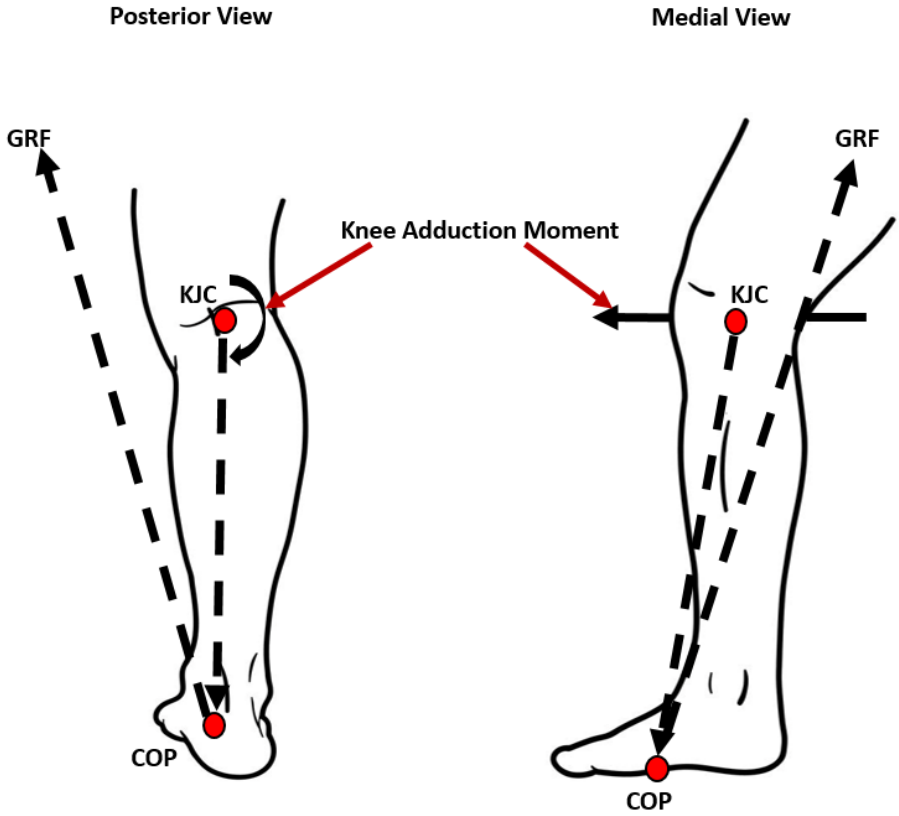

Figure 2.4. Posterior and medial views of vectors involved in calculating the KAM. Knee Joint Centre (KJC) is defined as the as the midpoint of the medial and lateral epicondyle markers. Centre of Pressure (COP) is the point on the ground where the ground reaction force (GRF) vector is considered to originate.

The KAM typically exhibits two peaks, the first peak during early stance when there is contralateral toe off and the second peak during late stance prior to contralateral heel strike (Figure 1.1) [121]. The magnitude of the first peak (usually largest) has been correlated with the severity [122], pain [123], and rate of progression of knee OA [3]. Although the main strategy of gait retraining has been to reduce the KAM, a recent study suggests that reducing the KAM does not necessarily reduce the medial contact forces if the knee flexion moment increases [124]. It is important to recognise that the KAM is just one of six inverse dynamic loads that can influence forces at the knee.

2.2.1 Gait Parameter Modifications

This section details the gait parameters that have been modified in literature in the attempt to reduce the KAM. An in-depth review has been written by Simic et al. [125].

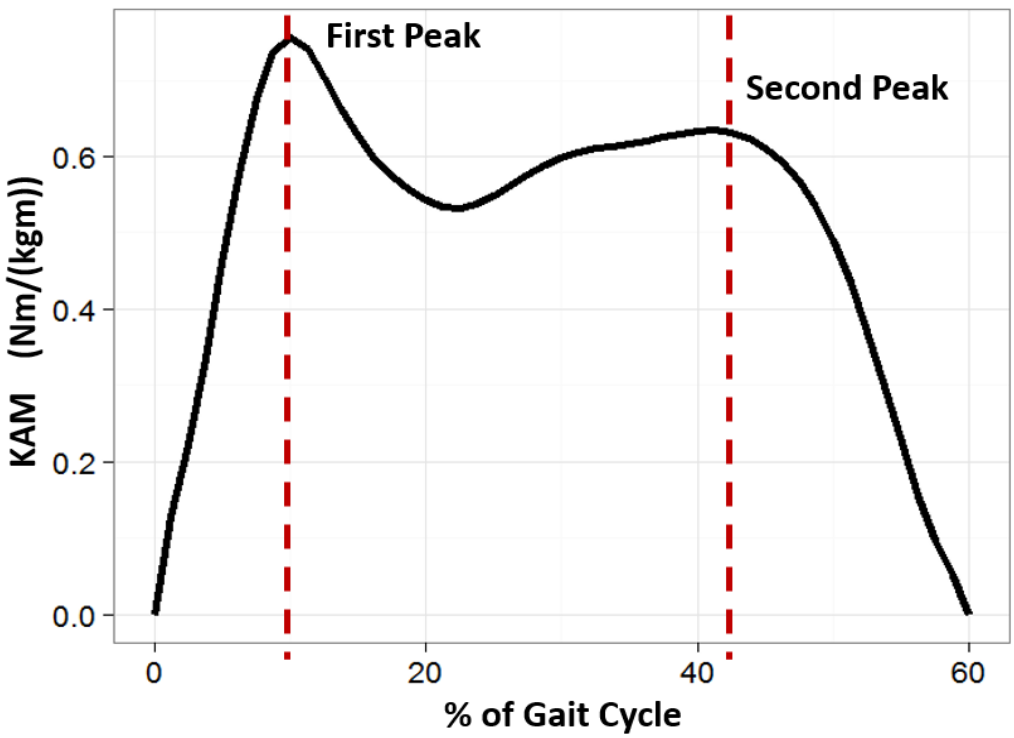

Figure 2.5. Example of the knee adduction moment during the gait cycle, showing the first and second peaks.

2.2.1.1 Foot Progression Angle (FPA)

The foot progression angle has been a commonly applied gait modification and can be defined as the angle of the foot (calcaneous to head of second metatarsal) with respect to the direction of forward walking progression (Figure 1.1). Two modifications are possible with the FPA, either toeing out or toeing in. Toeing out is defined as rotating the foot externally with respect to the baseline angle, and toeing in is thus internal rotation of the foot.

Most studies that have explored toeing out as a gait modification have seen decreases in the second KAM peak [126-131] but inconsistent results in the first KAM peak in early stance [132]. During toe-out gait, the first peak has been shown to decrease [133, 134], increase [126, 135], or have no significant change at all [129]. When the foot is externally rotated, the shift in the knee joint axis converts a portion of the KAM into a knee flexion moment, contributing to the reduction of the first KAM peak [133]. This shift causes the GRF to be more posterior than medial with respect to the knee joint axis. The mechanism for the reduction in the second peak, however, occurs due to the decreased magnitude of the moment arm due to the lateral shift of the centre of pressure of the foot [133]. This lateral shift in the centre of

pressure is only significant in late stance because during early stance, the COP is located closer to the heel, which is far less affected by external rotation.

Toe-in gait is another gait modification that can be performed to influence the KAM. There is no consensus in the effect of toeing-in on the first and second KAM peaks. Previous literature on this gait modification have reported decreases [127, 136], increases [134], and no change [127, 137] in the first KAM peak and, likewise, both increases [134, 137] and no change [127, 136] in the second peak. Toe-in gait reduces the KAM by shifting the knee joint centre medially. This causes a reduction in the moment arm of the GRF during early stance but is offset by the centre of pressure shifting medially during late stance. The positioning of both the knee joint centre and centre of pressure of the foot are important for reducing the moment arm and inconsistencies in reductions of the KAM could be due to the centre of pressure moving more medially than the knee joint centre [136].

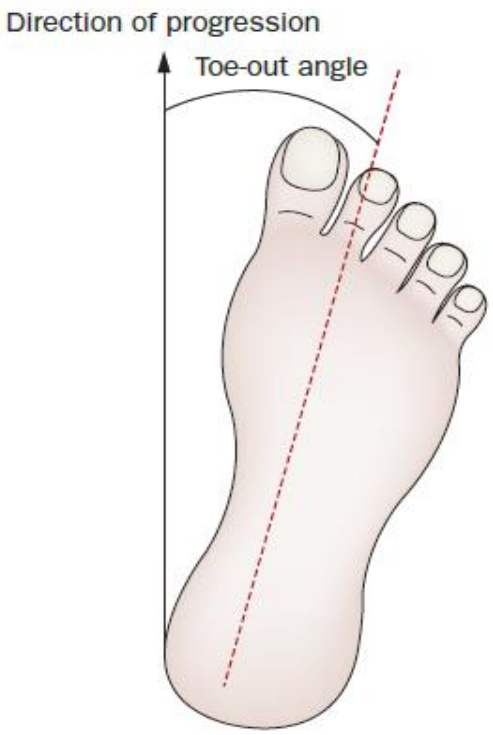

Figure 2.6. Toe-out foot progression angle with respect to the direction of progression. Image taken from [134].

2.2.1.2 Trunk Sway

Laterally swaying the trunk towards the side of the weight bearing foot shifts the centre of mass towards the knee joint centre, as shown in Figure 1.1. The reduction of the moment arm thus

results in a decrease in the KAM. Previous studies have shown increasing trunk sway to be effective in reducing both KAM peaks [4, 138]. Despite the effectiveness of lateral trunk sway as a gait modification for reducing the KAM, potential difficulties in adopting this change include an increased risk of falling associated with excessive upper body sway, its dependency on the individual's hip abductor muscle strength [138], and that increasing trunk may appear to be too unnatural and deterring individuals from adopting it.

2.2.1.3 Medialized Knee

Adopting a medialized knee gait involves shifting the knee joint centre medially during walking, which in turn reduces the moment arm from the GRF. Studies have shown medializing the knee as a possible gait modification for reducing both first [50] and second KAM peaks [131]. Although being a subtle gait modification that can be easily adopted, one caveat that has been identified is the knee flexion moment tends to increase as well [50], and thus may actually increase the medial contact force despite the decrease in the KAM [124].

2.2.1.4 Step Width

By increasing the width between each successive step, it is possible to shorten the lever arm associated with the GRF and thus reduce the KAM [128, 130]. It has been reported that during stair descent, increasing step width (SW) decreased both the first and second KAM peaks in healthy individuals [139] but not in patients with knee OA [140]. During stair ascent, however, both healthy and knee OA patients experienced reductions in the first and second KAM peaks [141]. Experimental studies during level walking have not been consistent in whether increasing SW decreases the KAM. Individuals in one study were reported to have decreased their first KAM peak by adopting a narrowed gait but this reduction was only observed in the non-dominant limb [142], whereas Favre et al. [143] observed increasing SW lead to reductions in the first KAM peak. It has been suggested that inconsistencies in the literature may be due to secondary changes that result from SW gait changes that are otherwise not accounted for [143].

2.2.1.5 Other Parameters

Other gait modifications that have been explored to reduce the KAM include altering gait speed [144-150], hip internal rotation [151], transferring weight to the medial side of the foot [5], increased knee flexion [152], and reduced stride length [153]. The majority of studies that

explored the influence of gait speed and stride length on peak KAMs did not observe any significant changes. Although increasing knee flexion resulted in higher peak KAMs in healthy individuals [152], no association was found in patients with symptomatic medial tibiofemoral knee OA [154]. Hip internal rotation [151] and medial weight shift of the foot [5] was observed to be effective in reducing the peak KAM. Despite the many modifiable gait parameters that exist, it is possible that many are correlated (for example tibia and foot progression angle through dynamic coupling [65]) and thus a gait retraining scheme should only need a subset of these.

2.2.2 Wearable Haptic Devices for Gait Retraining

Having introduced the haptic modalities that can be used as a directional feedback cues and gait parameters that have previously been modified for reducing the KAM; it is fitting to discuss the intersection of these two areas of research. Wearable haptic devices for gait retraining have been successfully integrated into real-time gait retraining schemes for many years for which Shull et al. provide a comprehensive review [155].

The gait parameter which has had the most interest with wearable haptics during gait retraining has been trunk sway and tilt [4, 7-9, 11-13]. The majority of studies have focused on assisting standing balance and improving stability for fall prevention through the use of vibrotactile head mounts [7-10] and vibrotactile vests [11, 13]. The only application of wearable haptic feedback on modifying trunk sway for reducing the KAM was a novel rotational skin stretch device on the back [156] to inform an individual on the amount of his or her trunk sway during gait retraining [4].

The remaining wearable haptic feedback devices for gait retraining with the goal of reducing the KAM are scarce. Wheeler et al. used binary vibrotactile feedback on the forearm to alert the individuals when their KAM was over a certain threshold. The individual was given the freedom to modify their gait as they saw fit [6]. Dowling et al. attached a vibration motor to the shoe, notifying the individual when his or her foot centre of pressure needed to be more medial [5]. Shull et al. attached vibration motors to the foot and tibia of an individual, successfully retraining the foot progression angle and tibia angle [4].

Despite the plethora of gait modifications which exist to reduce the KAM, relatively few wearable haptic devices have been developed and implemented to retrain them. This limits the

majority of gait retraining schemes to being confined to clinical environments. It has also illuminated the need for research in new wearable haptic devices and also their application towards modifiable gait parameters for the purpose of reducing the KAM in real-time.

Chapter 3: Combining Multiple Haptic Modalities for a Feedback Device

Effectiveness, comfort/acceptance, and portability are three requirements which have been identified as necessary elements for the adoption of wearable assistive devices. Despite the subjectivity which accompanies some of these requirements, discussions within the literature give insight as to what would accomplish these objectives. The focus of this section will predominantly be on how combining haptic modalities can contribute to those requirements.

The majority of gait modifications which show promise in reducing the KAM have focused on altering the kinematics of the lower limb with the exception of increasing trunk sway. When you consider the requirements of comfort and acceptance for a device which is intended to be used outside of clinical settings, trunk sway does not seem to be a suitable modification parameter. Increasing trunk sway has previously been identified as appearing unnatural [50, 157] and more difficult to modify as compared with changing the FPA [157]. Thus, it seems gait retraining should focus on kinematic modifications of the lower limb.

Wearable haptic feedback devices which alter lower limb gait have successfully used binary (on/off) vibrotactile cues [4, 5]. As it currently stands, altering multiple parameters require individual devices attached to the relevant segments e.g. on the foot for FPA and on the shank for tibial angle [4]. This presented an opportunity to further the portability of existing wearable feedback systems by combining multiple haptic cues into a single feedback device (see illustrated in Figure 1.1). For two gait parameters to be trained, two different patterns or cues have to be taught to the individual and be robust enough that perception accuracy is high during walking. Previous work has shown that vibrotactile pattern recognition accuracy vary depending on body loci with the torso yielding higher accuracies than the forearm [158].

Previous work in user perception of vibrotactile feedback during walking and running have shown that using modalities which differ significantly from each other increases accuracy of perception [159]. In Chapters 4 and 5, I sought to investigate skin stretch and TAM as modalities which could be synergistically combined with binary vibrotactile feedback. In Chapter 6, the two haptic modalities, skin stretch and TAM are discussed, and justification is provided as to why the final multi-modal haptic device is a combination of TAM and binary vibrotactile feedback.

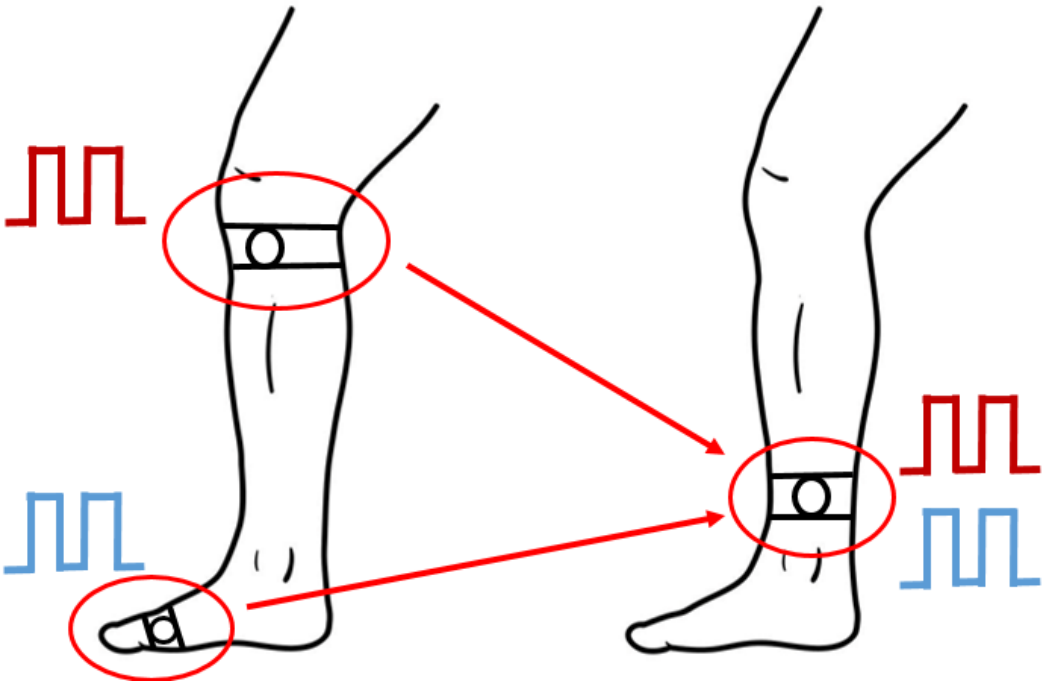

Figure 3.1. Part of the wearable haptic feedback system from Shull et al. [4] shown on the left. Proposed haptic device providing multiple haptic cues on the right.

Chapter 4: Exploring New Haptic Modalities for Lower Limb Feedback – Skin Stretch

Lower Extremity Lateral Skin Stretch Perception for Haptic Feedback

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Abstract—Tactile feedback in recent decades has allowed humans to receive information through technology beyond traditional visual and auditory senses. Lateral skin stretch has the potential to be a mode of tactile feedback, reliably enabling the perception of directional cues through the use of a single actuator. Experiments were conducted to explore sensitivity to skin stretch on nine locations on the human lower leg. Thirty-two stimuli were presented to individuals, exploring effects of displacement (from 0.2–2.0 mm) and speed (from 0.5–4.0 mm/s) on the perception of left and right directions. Higher accuracy came from stimuli having higher displacements and speeds. Three of the locations: soleus, calcaneal tendon (upper), and fibularis longus (lower) all had a mean accuracy of at least 85 percent and are suitable locations for a skin stretch tactile feedback device.

4.1 Introduction

Research in the area of haptic feedback has developed technology whereby humans can perceive information through skin sensations such as pressure, vibration, and stretch [160]. This has opened up new areas of innovation, some of which include novel human-computer interactions [161], proprioceptive feedback [162], and motion guidance [163].

The use of vibration as a modality dominates current wearable haptic feedback devices due to its high perceivability by humans, device compactness, and simple implementation using relatively low-power actuators [162]. Vibration, however, is not inherently directional, thus when it is used in a motion training task, multiple vibration actuators are required to signal directional information. Skin stretch has been proposed in recent studies as an alternative haptic feedback modality due to its ability to portray not only the magnitude of stimulation but also the direction in which it is stimulating using only a single actuator. Moreover, because skin stretch targets the slowly-adapting mechanoreceptors, it does not require a continuously varying stimulus like that of vibration [164]. This could not only prevent desensitization from occurring due to a continuous vibration sensation [80], but also reduce the number of actuators required in a haptic feedback device for motion training.

The authors involved have a particular interest in the field of lower limb motion guidance, especially the ability to guide an individual's lower extremities in the horizontal direction for parameters like stance width. This paper is therefore motivated by the fact that the capacity for skin stretch perception on the lower extremities is not yet known and thus need to be explored in order to determine whether skin stretch is a suitable modality for haptic feedback.

This study sought to quantify the accuracy perception of left and right skin stretch stimuli at different displacements and speeds. These experiments were conducted on nine different locations on participants' legs which were locations which the authors decided were practical locations for a wearable device.

4.2 Background

Tactile stimuli are sensed by a range of mechanoreceptors in the human skin, with different mechanoreceptors perceiving a different stimulus. The primary afferent for a vibration stimulus is the FAII, where the peak sensitivity occurs when the frequency of vibration of the skin is around 250Hz [165]. A lateral skin stretch stimulus, in contrast, is encoded mainly by the SAII afferents [166]. These afferents continue to fire if there is a static skin stretch stimulus present, whereas FAII requires a continuously changing stimulus for activation to take place.

Previous experiments with fingertips and forearms show that tangential displacements of the skin are more perceivable compared with normal displacements [101]. Direction detection thresholds are also lower for skin stretch compared with spatiotemporal stimuli (e.g.

blowing an air-jet or rolling an object across the skin which sequentially activates mechanoreceptors) [104]. Lower thresholds for lateral skin stretch could reduce the power consumption compared with the use of traditional vibration actuators. Experiments by Verrillo et al. [71] and Inglis et al. [102] showed older test participants having increased thresholds for high frequency stimuli, suggesting vibration may not be as effective with increasing age.

Existing skin stretch devices have been focused on the upper limbs, namely the fingers [109, 111, 112, 114-116, 118, 119, 167, 168], hands [117, 169] and forearms [108, 110, 113]. The exception to this has been the hydraulic stimulator for the lateral tibia used by Wasling et al. [120], who investigated cortical processing of tactile direction discrimination. The directional perception of lateral skin stretch for tactile applications on the lower limb holds much promise, but has yet to be explored.

4.3 Methods

A two-axis experimental setup (Figure 1.1) was constructed from two Velmex linear stages (XN10-0120-M02-71 and XN10-0080-M02-71) with a Vexta PK245-01AA stepper motor driving the horizontal linear stage. The vertical linear stage was manually controlled after measurements were made with a tape measure, depending on which location was being tested. An Interface SM-50N force transducer was attached during the pilot experiment to the vertical linear stage to measure the normal contact force with a IBM ThinkPad TrackPoint tactor on the end. During the main experimental phase, a Loadstar RAPG-001M-A load cell was used to measure the peak shear forces during skin stretch.

The stepper motor was driven with a Cytron SB02D motor driver and load cells were interfaced with a Loadstar DI-1000. The hardware used for data acquisition was a National Instruments USB-6351 DAQ device with a laptop running LabVIEW. It must be noted that accelerations and decelerations caused by the stepper motor inevitably contain high frequency components, which are very difficult to remove, and will to a certain extent activate the FAII mechanoreceptors as well during stimulation.

A brief pilot experiment was conducted to determine what normal contact force was required to enable the tactor and skin to not slip during stimulation and it was confirmed that 0.25 N as reported by Gleeson et al. [107], was sufficient. In the experiments, no external physical constraint was placed on the test participant’s leg. The participant was told to relax

and keep their leg “as still as possible” during the stimulations. Adhesive tape was used to reduce slipping during stimulation as performed in previous experiments [120, 170, 171].

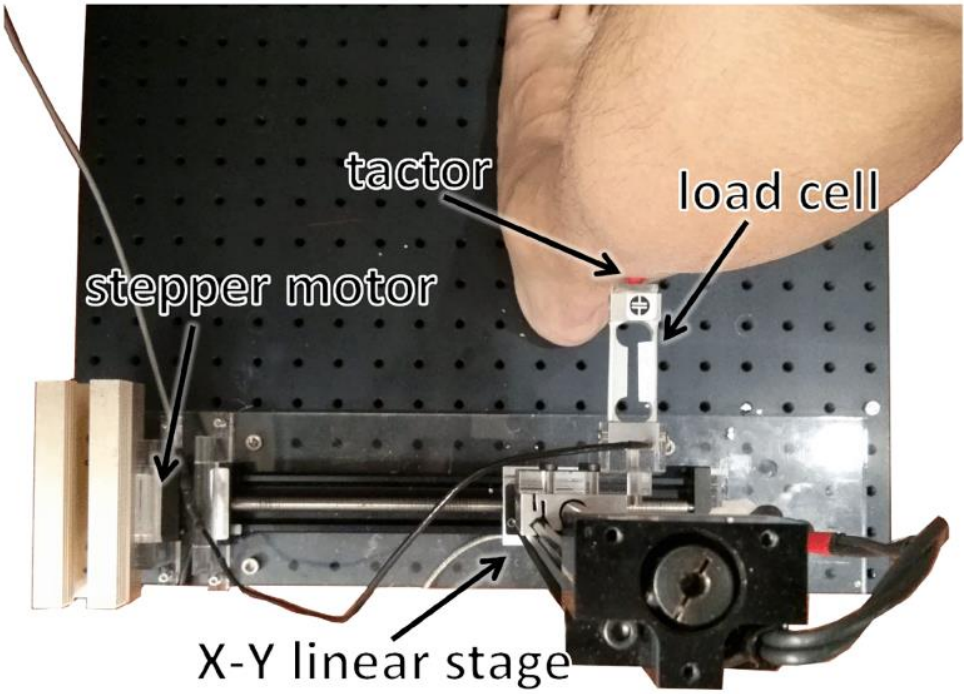


Figure 4.1. The two axis linear stages rendering lateral skin stretch in the horizontal direction. The Loadstar RAPG-001M-A load cell is shown.

In this experiment, we wanted to determine suitable locations on the lower limb to place a skin stretch device and also what the speed and displacement requirements of such a device would be. Nine locations were chosen (Figure 1.1).

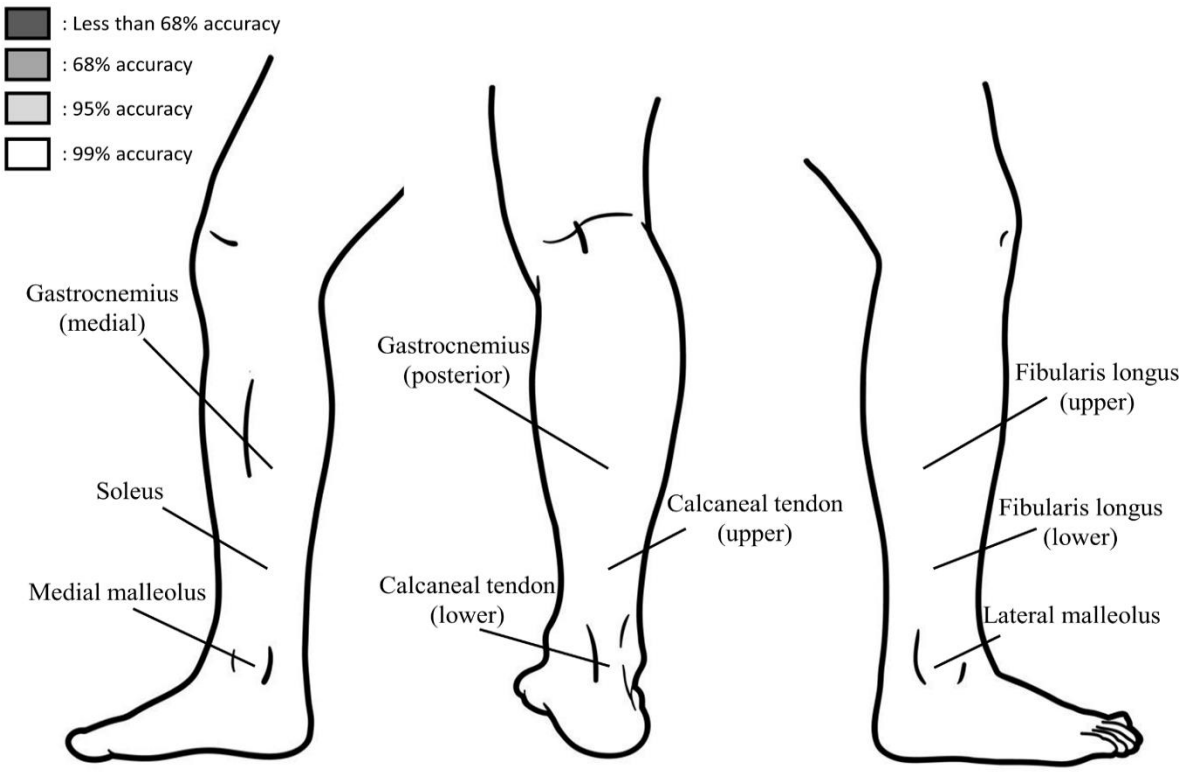
Eight test participants (six males and two females, range 20-30 years) volunteered to participate in this experiment and signed an informed consent form in accordance with the ethics committee of the University of Auckland. Each participant was seated in a chair with one of the nine locations of interest attached to the tactor via thin double-sided tape. Each set of experiments consisted of 32 different stimuli based on the four displacements of 0.2, 0.5, 1.0 and 2.0 mm, four speeds of 0.5, 1.0, 2.0 and 4.0 mm/s and the two directions (left and right) of interest. The left and right directions were defined with respect to an external observer facing the participant. For each of the nine locations, each participant completed five test sets of the 32 stimuli. Each test set averaged four minutes and participants were given rest periods after five sets (taking approximately 20 minutes) were completed. Participants had the choice of whether they wanted to continue with a different location after the rest period or if they wanted to continue with the next location the next day. The 32 different stimuli at the nine locations

were randomized in each test set to avoid pattern guessing. A familiarization test ensured the participant knew what to expect. Headphones playing background music were worn to mask any noise of the stepper motor.

Following each stimulus, the participant indicated whether the skin stretch was to the left, right, or 'not sure' if they were not sure. Four possible outcomes occur when a stimulus is presented to the participant: they did not feel anything, felt something but were unsure of the direction, felt something but reported the direction incorrectly, or felt something and reported the direction correctly. The addition of the 'not sure' option means that participants do not have to guess if they are not certain which direction the stimulus is going towards allowing for accuracies lower than 0.5, which is the usual chance level accuracy if the two-alternative forced choice method were used. Stimuli that were not perceived correctly can be categorized as being either outcomes 1 and 2 or outcome 3 (mentioned above), giving the ability to determine whether the incorrect perception was due to the participant not being able to feel the direction or whether it was because there was some physiological factor causing the opposite direction to be felt instead. After the answer was logged into the LabVIEW program, the tactor returned to its original starting position before the next stimulus was presented. Previous vibrotactile experiments suggested a 300 ms pause is sufficient in preventing any stimulus masking [172].

A confusion (or error) matrix (Figure 1.1) was constructed for each location by pooling data from all the test participants. The matrices display the number of stimuli accurately perceived, for each location out of the total number of stimuli rendered.

Although this experiment only tested the outbound stimulus, users commented that the return stimulus back to the original position reinforced the direction cue, which was also observed by Gleeson et al. [107].



Gastrocnemius (medial)

Distance (mm)	Speed (mm/s) 0.5	1	2	4
0.2	0.31±0.12	0.31±0.14	0.53±0.17	0.94±0.08
0.5	0.57±0.10	0.69±0.15	0.77±0.15	0.94±0.04
1	0.70±0.15	0.89±0.10	0.93±0.08	1.00±0.00
2	0.86±0.07	0.81±0.21	0.93±0.08	0.97±0.04

Gastrocnemius (posterior)

Distance (mm)	Speed (mm/s) 0.5	1	2	4
0.2	0.49±0.21	0.63±0.16	0.60±0.22	1.00±0.00
0.5	0.69±0.11	0.79±0.18	0.89±0.08	0.93±0.06
1	0.84±0.08	0.87±0.07	0.96±0.06	0.99±0.03
2	0.89±0.05	0.87±0.15	0.93±0.11	1.00±0.00

Fibularis longus (upper)

Distance (mm)	Speed (mm/s) 0.5	1	2	4
0.2	0.41±0.15	0.49±0.14	0.69±0.13	1.00±0.00
0.5	0.59±0.17	0.84±0.11	0.93±0.06	0.96±0.08
1	0.86±0.10	0.97±0.04	0.97±0.04	1.00±0.00
2	0.94±0.06	1.00±0.00	1.00±0.00	0.97±0.04

Soleus

Distance (mm)	Speed (mm/s) 0.5	1	2	4
0.2	0.43±0.21	0.63±0.22	0.74±0.17	0.97±0.06
0.5	0.64±0.21	0.87±0.10	0.94±0.07	1.00±0.00
1	0.74±0.12	0.89±0.13	0.94±0.06	1.00±0.00
2	0.83±0.16	0.94±0.08	0.91±0.11	0.99±0.03

Calcaneal tendon (upper)

Distance (mm)	Speed (mm/s) 0.5	1	2	4
0.2	0.30±0.10	0.23±0.09	0.43±0.14	0.98±0.03
0.5	0.58±0.14	0.57±0.19	0.72±0.15	0.83±0.13
1	0.57±0.19	0.76±0.19	0.89±0.09	0.93±0.06
2	0.79±0.12	0.83±0.13	0.97±0.03	0.99±0.02

Fibularis longus (lower)

Distance (mm)	Speed (mm/s) 0.5	1	2	4
0.2	0.43±0.23	0.73±0.15	0.84±0.10	1.00±0.00
0.5	0.71±0.23	0.84±0.15	0.93±0.08	1.00±0.00
1	0.80±0.18	0.91±0.08	0.96±0.06	0.99±0.03
2	0.87±0.13	0.96±0.04	0.99±0.03	1.00±0.00

Medial malleolus

Distance (mm)	Speed (mm/s) 0.5	1	2	4
0.2	0.37±0.18	0.57±0.10	0.70±0.16	0.99±0.03
0.5	0.57±0.18	0.74±0.08	0.93±0.07	0.93±0.07
1	0.76±0.07	0.89±0.07	0.94±0.06	0.91±0.03
2	0.77±0.07	0.93±0.06	0.94±0.08	0.99±0.03

Calcaneal tendon (lower)

Distance (mm)	Speed (mm/s) 0.5	1	2	4
0.2	0.21±0.12	0.50±0.16	0.57±0.19	0.97±0.04
0.5	0.47±0.15	0.54±0.19	0.70±0.14	0.87±0.09
1	0.64±0.15	0.60±0.18	0.74±0.13	0.87±0.11
2	0.66±0.14	0.70±0.15	0.81±0.12	0.94±0.07

Lateral malleolus

Distance (mm)	Speed (mm/s) 0.5	1	2	4
0.2	0.27±0.19	0.54±0.18	0.83±0.11	1.00±0.00
0.5	0.59±0.17	0.81±0.12	0.91±0.07	0.99±0.03
1	0.70±0.04	0.89±0.07	0.93±0.07	1.00±0.00
2	0.81±0.10	0.93±0.07	0.99±0.03	0.99±0.03

Figure 4.2. Diagram showing the nine locations tested along with tables showing their corresponding mean direction perception accuracy, and 95 percent confidence intervals for each stimulus. Locations shown are: gastrocnemius (medial) which is 240 mm from the bottom of the heel, soleus which is approximately the same location of the fibularis longus (lower) reflected onto the medial side, the outer most point of the medial malleolus, gastrocnemius (posterior) which is located 240 mm from the bottom of the heel, calcaneal tendon (upper) is located at the same height as the soleus, and fibularis longus (lower) just on the posterior side, and the calcaneal tendon (lower) which is located at the same height as the outer most point of the lateral malleolus on the tendon, fibularis longus (upper) which is 240 mm from the bottom of the heel, fibularis longus (lower) which is half way between the bottom of the heel, and the edge of the calcaneal tendon, and the gastrocnemius, and the outer most point of the lateral malleolus. All measurements were made relative to a fixed anatomical point, the heel, similar to previous lateral skin stretch studies by Wasling et al. [120] and Lundblad et al. [171]. The shading within tables give an indication of whether the accuracy is less than 68 percent, at least 68 percent, at least 95 percent, or at least 99 percent (as seen in the legend).

Direction Rendered	Direction perceived: Right	Left	Not Sure
Right	a	b	c
Left	d	e	f

$$\text{Accuracy rate} = \frac{a+e}{a+b+c+d+e+f}$$

$$\text{“Not Sure” Perception rate (NR)} = \frac{c+f}{a+b+c+d+e+f}$$

Figure 4.3. A confusion (or error) matrix was constructed to compile all the stimuli for each location where an overall accuracy rate, and the rate at which the participant didn't feel or was unsure of the direction of stimulus can be calculated. Accuracy rates from these matrices are used to determine locations suitable for a skin stretch device.

4.4 Results

Results for each location were compiled from all participants into a matrix showing the accuracy of each stimulus and their corresponding 95 percent confidence intervals (Figure 1.1). In general, for all locations, larger displacements and speeds improved the accuracy (Figure 1.1), similar to prior research by Olausson and Norrsell [106].

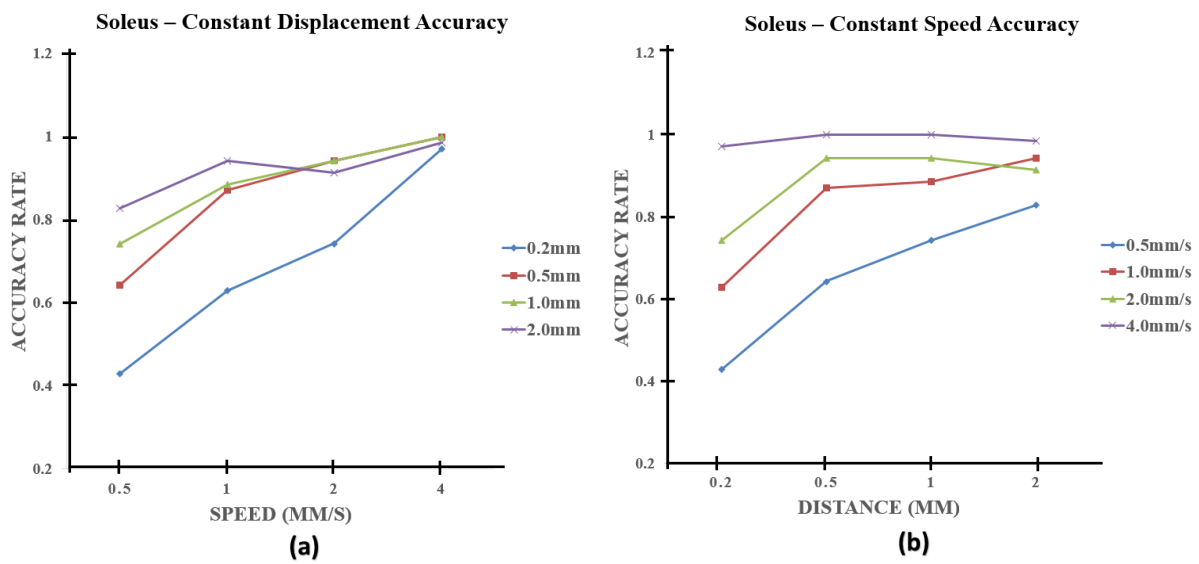


Figure 4.4. Graphs of accuracy rate for each individual stimulus with (a) showing constant displacement, and (b) constant speed. Trends from both plots showed that increasing the displacement and speed both improved the accuracy.

An overall accuracy rate was calculated based on the confusion matrices. A one-way ANOVA with a Bonferroni corrected post hoc test was performed with the different locations being the factor of interest, $F(8, 56) = 51$, $p < 0.001$. The lower calcaneal tendon (Location 7) was the least sensitive location with an accuracy rate of 0.67 ± 0.03 ($p < 0.01$ compared to other locations). The most sensitive locations were the upper and lower fibularis longus (Locations 2 and 5, 0.86 ± 0.01 and 0.85 ± 0.01, respectively), upper calcaneal tendon (Location 4, 0.87 ± 0.01), soleus (Location 6, 0.85 ± 0.02), and the lateral malleolus (Location 9, 0.85 ± 0.01). There was no statistical difference in accuracy rates between these most sensitive locations ($p > 0.05$).

With the addition of the "not sure" option, the rate at which the participant did not feel or was unsure of the direction of stimulus (NR) was also calculated. An overall NR was calculated using the confusion matrices. A one-way ANOVA with a Bonferroni corrected post hoc test was performed with the different locations being the factor of interest, $F(8, 56) = 44$, $p < 0.0001$. The location with the highest NR was the posterior gastrocnemius at 0.20 ± 0.04, which was higher than all other locations ($p < 0.05$). The upper calcaneal tendon had the lowest NR at 0.08 ± 0.02. This difference was also shown to be statistically significant compared with all other locations ($p < 0.05$).

4.5 Discussion

The closest data for direct comparison from the literature were by Wasling et al. [120] and Lundblad et al. [171], where distal and proximal direction skin stretch was performed near the fibularis longus (upper) location. Similar to our experiment, their tactor was glued to the leg and stimulations of 2 mm at a speed of about 1 mm/s were performed. They found that a shear force of 1.4±0.2 N was required for the stimulation, which was in range of our required force of 1.71 ± 0.14 N for that stimulus. They noted accuracy rates of 0.86 ± 0.16, 0.92 ± 0.08 and 0.94, as they performed the experiment on three separate occasions with the third experiment only having one participant. This was slightly lower than what we observed at the upper fibularis longus, where we had an accuracy rate of 1. This could be due to the particular location having a greater sensitivity in the posterior and anterior directions as opposed to the distal and proximal directions. Previous experiments have noted directional biases of SAII receptors [99, 173]. Further experiments by Lundblad et al. [171] using a rolling wheel on the thigh as a spatiotemporal stimulation, having stimulation distance of 35 mm and speed of 20

mm/s, yielded an accuracy of approximately 0.98 ± 0.01. This supports the notion that the leg is more sensitive to skin stretch than spatiotemporal stimulations, requiring significantly lower stimuli in terms of magnitude and yielding higher accuracy.

Olausson et al. [170] performed tactile perception experiments for the forearm and using a stimulus displacement of 0.242 mm at a speed of 30 mm/s they obtained an accuracy of 0.77 ± 0.06. This implies that the locations we tested on the lower limb are more sensitive than on the forearm, with even our least sensitive location (lower calcaneal tendon) having an accuracy rate of 0.97 ± 0.04 when we presented the participants with a stimulus of 0.2 mm displacement and a speed of 4 mm/s. Upon comparing our data with the fingertip study from Gleeson et al. [107], it is clear that the skin on the fingertip is more sensitive than the lower limb. At a displacement of 0.2 mm and a speed of 4 mm/s, the accuracy rates of any locations that were tested on leg, were approximately equal to or better than the accuracy on the fingertip. When the speed was reduced to 2 mm/s, the difference becomes apparent, with the fingertip having a minimal change in accuracy rate of 0.96 ± 0.03 from 0.95 ± 0.03 whereas the most sensitive area of the leg, the lower section of the Fibularis longus, had a reduced accuracy of 0.84 ± 0.10 from 1. Norrsell [105] showed that, when applying directional spatiotemporal stimulation on the forearm, a displacement of 17.5 mm was required with an air-jet before the participant could identify the direction, confirming that lateral skin displacement is more sensitive than frictionless spatiotemporal stimulation. This was also observed by Gould [104], when spatiotemporal stimulation with an air-jet was compared with lateral skin displacement.

It is not surprising that the locations with the highest NR, the posterior gastrocnemius, happened to be the location with one of the lowest accuracy rates. Likewise, it is also not surprising that the location with the lowest NR, the upper calcaneal tendon, was the location with the highest accuracy rate. It is, however, unclear why the difference between the NR of the posterior gastrocnemius was significantly different from the other locations with low accuracy rates and likewise, why the difference between the NR of the upper calcaneal tendon was statistically significant from locations with high accuracy rates. One possible reason for this difference could be the physiological differences in the distribution of the Pacinian corpuscles among the different locations, meaning that when smaller stimuli were presented, the vibrations of the stepper motors used in this experiment masked the small skin stretch sensations intended.

Direct comparisons of the perception of lateral skin stretch on different locations could not be made as the authors were not aware of any comparable data in the literature. Very little is currently known about the distribution and characteristics of SAII mechanoreceptors on the locations that we tested. Previous studies exploring SAII receptors on the lower limb have been performed on the thigh [174], dorsal skin of the foot (lateral peroneal nerve) [175, 176], sole of the foot [177], calf [178], and the sural nerve of the foot [179]. Trulsson's [179] study of the sural nerve of the foot was the only study which showed distributions of SAII receptors which coincided with locations close to the ones the authors tested. No SAII receptors were found in Location 7 but it was noted that all SAIIs could be activated by skin stretch outside the receptive field. The sensitivity of SAII has also been known to be dependent on the direction in which the skin stretch is being applied [103]. Further investigation, stretching the skin at the same locations but in different directions and using linear motors instead of stepper motors, would

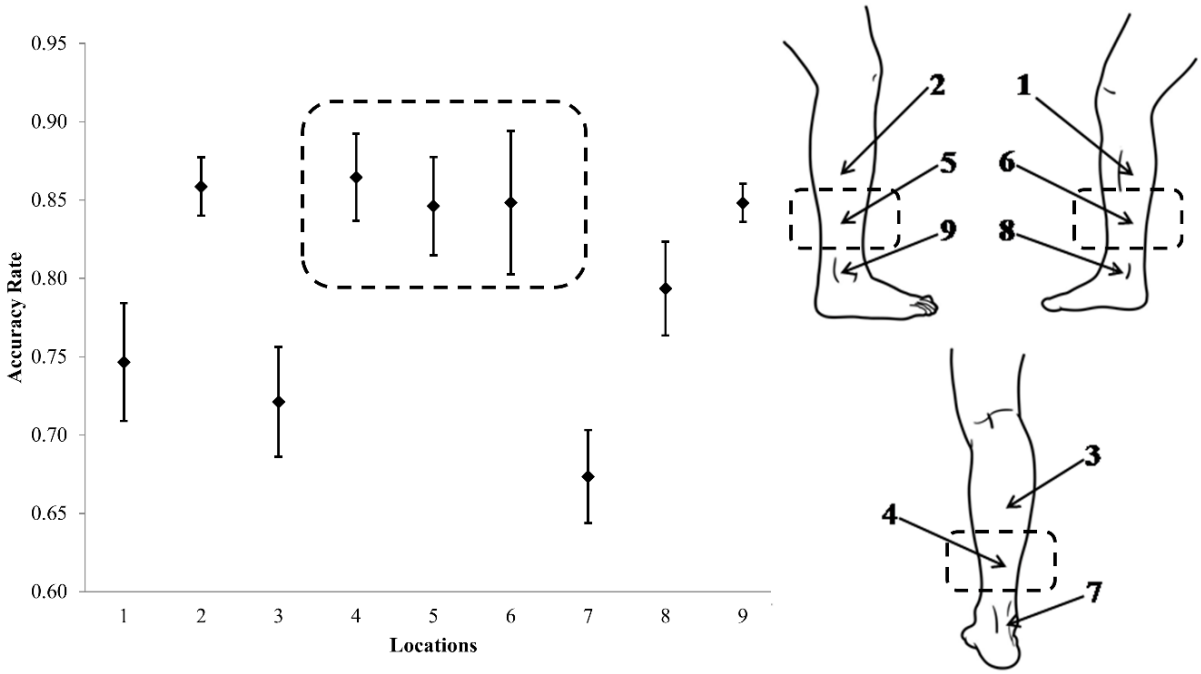


Figure 4.5. Graph of the accuracy rates for each location based on each of the eight individuals. Dashed boxes highlight the three locations: calcaneal tendon (upper), fibularis longus (lower), and soleus, as having among some of the highest sensitivity to skin stretch, and also convenient locations for a haptic feedback device.

be necessary in determining whether there indeed is a directional bias whilst minimizing any vibration masking.

Out of the five most sensitive locations, a possible region for a skin stretch device based on the grouping of certain locations emerged. This region encompassed the soleus, upper calcaneal tendon and the lower fibularis longus. These three locations all had a mean rate of at

least 0.85, and are conveniently located together allowing for a potential wearable haptic device above the ankle (Figure 1.1).

4.6 Conclusion

In this study, lateral skin stretch has shown to be a suitable method for providing direction tactile feedback to the lower limb when participants are static. Three locations on the lower limb, the soleus, upper calcaneal tendon, and the lower fibularis longus, have all displayed potential as locations for a skin stretch device. Participants were able to correctly perceive skin stretch directions as small as 0.2 mm at those locations with an accuracy rate of at least 0.91. This expands the possibilities for lower limb haptics for human-computer interactions, motion guidance, and proprioception. In order to extend these findings towards dynamic applications, e.g. gait retraining, experiments specific to the intended application would need to be performed.

Acknowledgements

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Chapter 5: Exploring New Haptic Modalities for Lower Limb Feedback – Tactile Apparent Movement

Tactile Apparent Movement as a Modality for Lower Limb Haptic Feedback

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Abstract - Wearable haptic technology has been shown to be effective for motion training and rehabilitation. However, one challenge is providing multiple intuitive tactile feedback during walking and hence new feedback methods need to be explored. Experiments were conducted to explore the use of tactile apparent movement on the lower extremity and its feasibility as a feedback modality. Optimal stimulus duration and inter-stimulus onset interval (ISOI) combinations were determined. We obtained the optimal mean ISOIs at six different stimulus durations (from 100 – 200 ms) and then measured the participants' left and right perception accuracy and response times when those stimuli were presented in a randomized trial during standing and walking. This study shows that apparent movement can be an effective feedback modality during walking, achieving accuracies of ~100% and low response times of < 1010 ms, given the optimal stimulus.

5.1 Introduction

In recent decades, haptic feedback research has given rise to new and innovative ways for motion training [163], augmenting proprioception [162], and human-computer interactions [161]. With most of the haptic feedback research being targeted towards the upper limbs and hands, much less has been explored for the lower limbs.

Human-computer interaction and motion training are areas where the authors have a deep interest, in particular that of lower limb haptics. Gait retraining is an example of an area where lower limb haptic feedback has shown promising results in slowing the progression of knee osteoarthritis [157]. Our long-term goal is to develop a portable and wearable device, which would allow for human-computer interactions and motion training for the lower limbs during walking. Current haptic feedback research for the lower extremities has focused on using binary vibratory sensations, which has been shown to be effective and intuitive. Using a single feedback modality, however, has the disadvantage of only being able to create one distinguishable stimuli and thus limiting the number of different interactions or movement parameters that can be recognized. Chen et al. have previously explored the use of lateral skin stretch as a feedback method inside a laboratory [180], and previous research has also been performed using saltation with limited success [181]. One modality that has not been extensively explored for the lower limb is that of tactile apparent movement (phi illusion). To investigate new potential feedback methods for a wearable lower limb haptic feedback device, we sought to determine the feasibility of tactile apparent movement as a feedback method for presenting directional (left and right) stimuli during walking.

The purpose of this study was to determine the parameters which illicit the optimal perception of apparent movement. “Optimal movement” in this context was defined as the best uninterrupted and continuous feeling of movement between stimuli [91]. The parameters that generated the optimal movement sensations were then evaluated based on directional perception accuracy and reaction time of the participants.

5.2 Background

Tactile apparent movement or phi illusion, is the sensation felt when two or more discrete loci on the skin are sequentially stimulated, causing a stroking illusion to be felt [90]. Early studies conducted experiments using simple indentations on the skin and discovered that perception improved when vibratory stimuli were presented instead of indentations. The main parameters influencing the perception of tactile apparent movement are the interstimulus onset interval (ISOI) and the stimulus duration [91]. The stimulus duration is the length of time the vibration is presented at any one time and the ISOI is time separating sequential vibrations as seen in Figure 1.1 below.

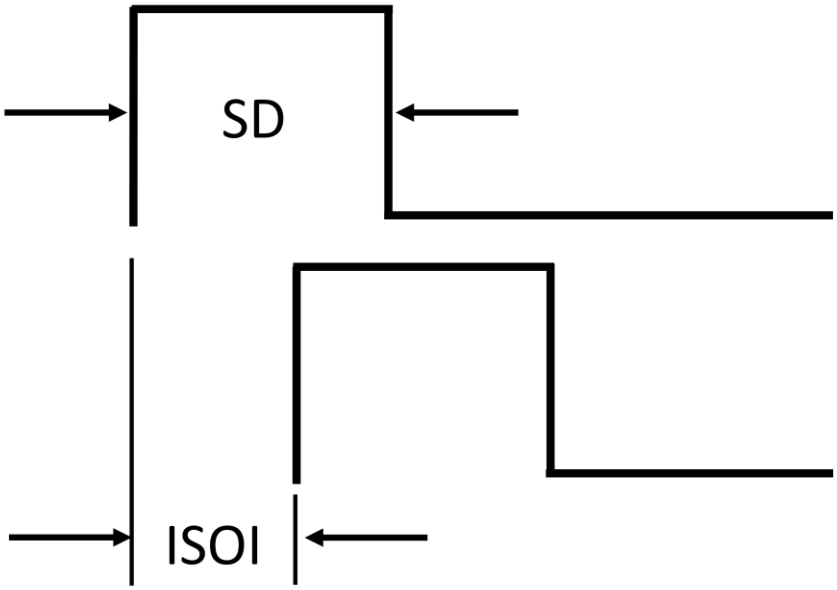

Figure 5.1. Signals of vibration motors being sequentially actuated. Stimulus duration (SD) and interstimulus onset interval (ISOI) representing the actuation time per motor and delay time between the start of each motor respectively.

Sherrick and Rogers [91] performed experiments on the ventral thigh with two stimulators with a vibration frequency of 150 Hz. Kirman [93] performed similar experiments on the finger pad with a vibration frequency of 100 Hz yielding results with a similar trend to Sherrick and Rogers. Kirman [94] reported that increasing the number of stimulators from two to four produced increased perceptions of apparent movement and resulted in shorter ISOIs. Interstimulator spacing and the shape of the stimulator have been observed to have little influence on the optimal ISOI [93].

Previous devices using tactile apparent movement include directional warning systems for drivers with actuators stimulating the posterior thighs [182], tactile display stimulating the back [183], tactile sleeves for forearm simulating different touch sensations [184], tactile actuators on the fingers [185], and tactile torso displays [186]. None of these previous applications address the use of apparent movement on areas suitable for a wearable lower limb haptic device while walking. Therefore, the aim of this study was to investigate the use of haptics to provide apparent movement on the lower limb during walking and identify the potential for this modality for gait retraining.

5.3 Methods

This study was divided into two experiments; the first experiment measured the ISOI and stimulus duration values that participants felt as having the “best apparent movement”, and the second evaluated the different “best” ISOI and stimulus duration combinations based on directional perception accuracy and reaction time.

5.3.1 Experimental Setup

Both experiments consisted of a LabVIEW program implemented on a National Instruments MyRIO device, which acted as a controller. The controller received a user adjustable voltage from a potentiometer, which determined the ISOI ranging from 2 – 296 ms. The controller sent I2C commands to the I2C multiplexer (TI TCA9548A) and the individual haptic drivers (TI DRV2604L) during initialization of the system and pulse-width modulated signals to the drivers during operating phase as shown in Figure 1.1. The haptic drivers measured the back-EMF of the eccentric rotating mass vibration motors (ERMs) minimizing the rise time to steady state during activation and deactivation of the motors. The ERMs had a diameter of 8 mm and a thickness of 2.5 mm. A LabVIEW PC client provided the graphical user interface (GUI) through which the participant interacted with the experiment whilst also logging the experimental data into a spreadsheet. The experiments were all conducted on a Force-Instrumented Treadmill (Bertec Corp, MA, USA). A force sensitive resistor was placed near the distal end of the 5th metatarsal on the plantar side of the participant’s right foot, which was used to detect when the foot was in stance phase. Tactile feedback was only provided when the participant’s right foot was in stance phase as this was felt to be most perceivable during our pilot test.

Four actuators were adhered to the skin using double sided tape to minimize slipping. The actuators were placed horizontally on the anterior hemisphere of the participants’ right legs with the actuators evenly distributed (Figure 1.1). The actuator was placed at a height halfway between the bottom of the heel and the edge of the calcaneal tendon and the gastrocnemius. This location was chosen based on our previous study, which determined the optimal location for sensing lateral skin stretch [180].

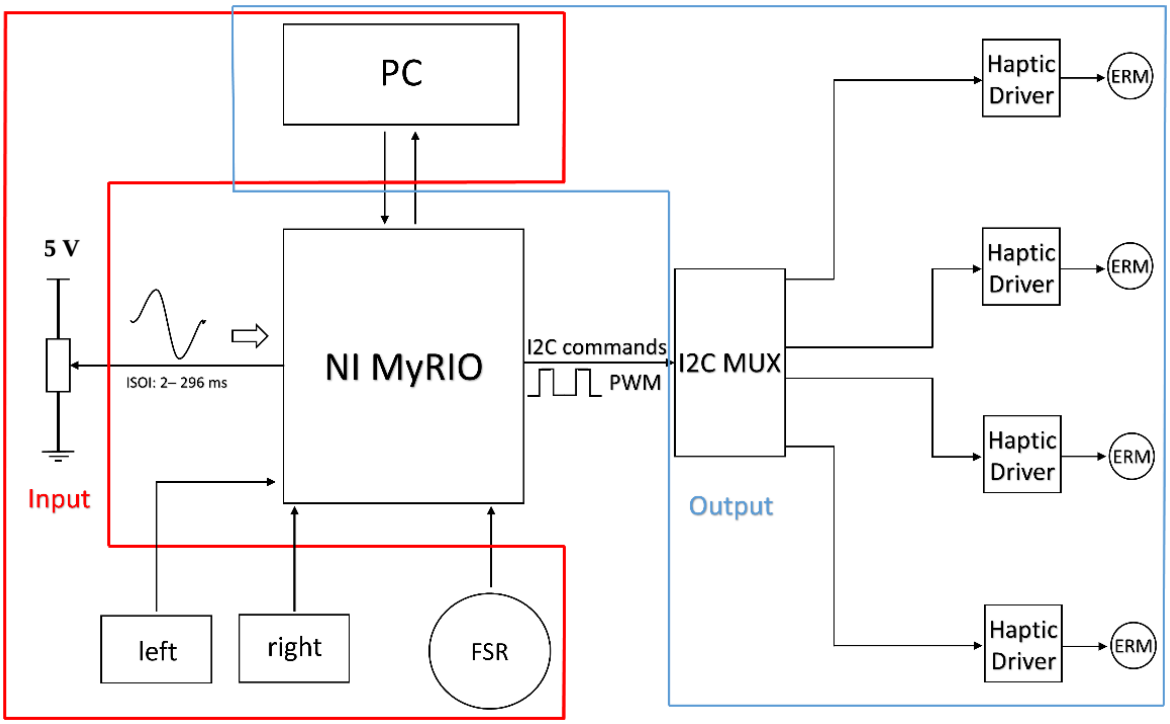

Figure 5.2. Experimental setup showing the inputs and outputs of the system. The participant adjusts the variable resistor which in turn determines the input voltage to the NI MyRIO controller which determines the interstimulus onset interval (ISOI). A force sensitive resistor (FSR) attached to the bottom of the participant's foot signals when the participant is in mid stance. Signals are sent to the eccentric rotating mass (ERM) motors during mid stance.

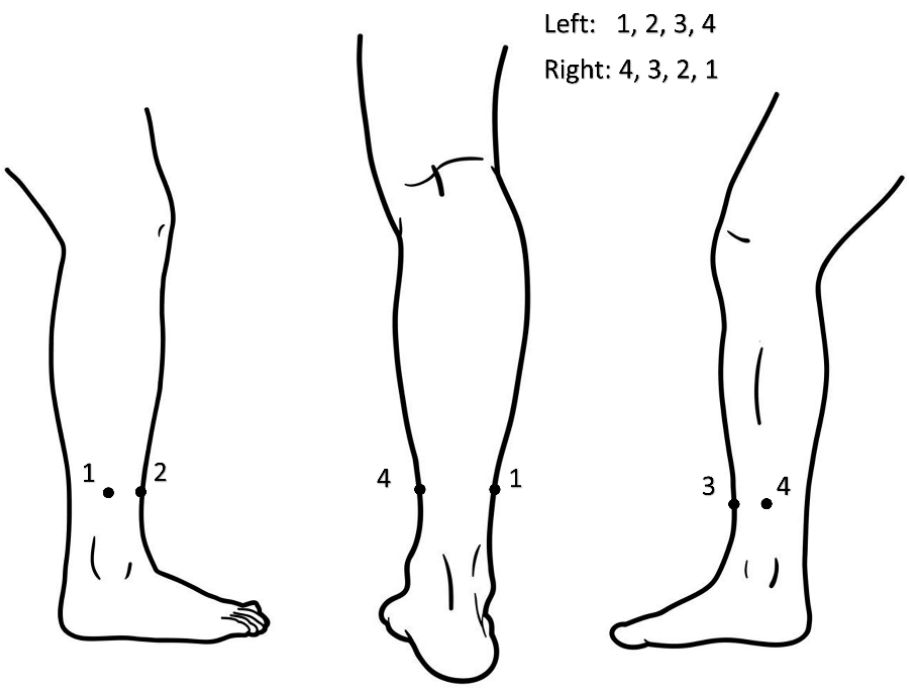

Figure 5.3. Lateral, posterior, and medial views of the right leg, with the black dots and numbers showing the positions of the vibration motors. Directions left and right are shown by the order of actuation of the motors.

5.3.2 Experiment 1: Mean Interstimulus Onset Interval and Stimulus Duration Combinations During Standing and Walking

Ten healthy participants aged from 23 to 33 years took part in this experiment of which there were 9 males and 1 female. Participants were recruited from within the university and qualified for the study as long as they did not have any self-reported neurological or lower limb disorders. Prior to testing, participants provided their informed consent to participate, to comply with the ethics committee of the University of Auckland. The aim of this experiment was to determine at which ISOI participants had the “best” perception of tactile apparent movement when the motors were actuated at different stimulus durations of 100, 120, 140, 160, 180, and 200 ms. The lower range of 100 ms was used as the authors felt that this was the shortest duration where the motors were still perceivable during walking. It was reported by previous studies that stimulus durations above 200 ms did not improve the perception of apparent movement and hence this was used as the upper limit [93]. An example of the “best movement" was presented to the participants using an ISOI value of 100 ms at the stimulus duration of 180 ms which was verified by the authors to be one which showed good tactile apparent movement. Each participant adjusted the potentiometer which varied the ISOI from 0 to 297 ms until they found a value which gave the desired stimulus.

The experiment was broken down in two parts, a standing and a walking experiment. Half the participants performed the standing experiment first before the walking and the other half performed it the other way round to minimize any learning effects. During both the standing and walking experiments, the stimulus duration and direction was also randomly presented to the participants. Six different stimulus durations and two different directions provided a total of 12 trials for each standing and walking experiment. During standing, each stimuli was presented with a 300ms rest in between to remove any temporal effects [172]. The walking experiments were performed at a treadmill speed of 1.2 m/s, which the participants felt was a comfortable walking speed and similar to their normal walking speed. Each stimulus was presented only once during each stance phase of the right foot. A force sensitive resistor was taped to the bottom of the foot near the head of the fifth metatarsal of the right leg. The maximum ISOI was also limited in software to reflect finite time in which the foot was in stance phase i.e. the stimulation would finish before the end of stance phase for a more consistent perception of tactile apparent movement.

Pairwise t-tests between left and right directions at different stimulus durations during standing and walking were performed ($\alpha = 0.05$) in R to determine whether or not the direction of stimulus had a significant effect on the ISOI.

5.3.3 Experiment 2: Accuracy and Response Time of Optimal Stimuli Combinations During Standing and Walking

Nine male individuals aged from 22 to 33 years took part in this experiment of which four had been individuals in the previous experiment. The aim of this experiment was to determine the direction perception accuracy and the response time when participants were presented with a random optimal mean ISOI and stimulus duration combination determined from experiment 1. Results from experiment 1 showed that there was not a direction bias. Therefore, the same ISOI values could be used for stimuli going left and right. In total, each stimuli combination was presented 20 times, half going left and half going right. With six different stimuli combination this resulted in 120 randomly presented stimuli. As with experiment 1, a crossover design study was implemented. When a participant was presented with a stimulus they would press either the left or right push button depending on whether they perceived it as going left or right, and the response time was recorded in each instance. Accuracy was determined by the number of correctly identified stimuli over the total number of stimuli for that particular combination (with total being 20) for both standing and walking. The mean response time for each participant was found for each of the six optimal stimuli for both standing and walking. The total session for each participant took no more than 15 minutes.

5.4 Results

5.4.1 Experiment 1

Standing and walking results of all the individuals were compiled showing the mean ISOI at the corresponding stimulus durations and their respective 95 percent confidence intervals (Figure 1.2 and Figure 1.1).

Pairwise comparisons within each stimulus duration comparing the left and right ISOI values using Bonferroni correction showed that there was no difference in the mean ISOI values between directions for both standing and walking activities, $p > 0.05$. A combined overall mean ISOI at each stimulus duration disregarding direction was thus calculated (Figure 1.1 and Table 1.1).

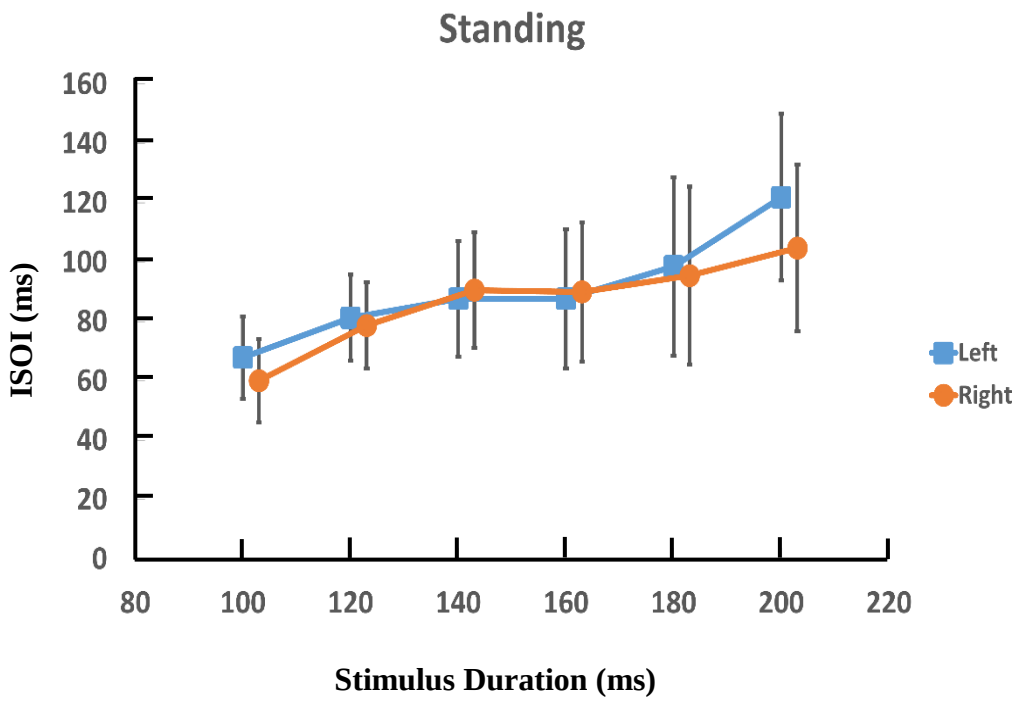

Figure 5.4. The mean and 95% confidence intervals for interstimulus onset interval (ISOI) values giving the "best apparent movement" during standing at different stimulus durations for left and right directional cues.

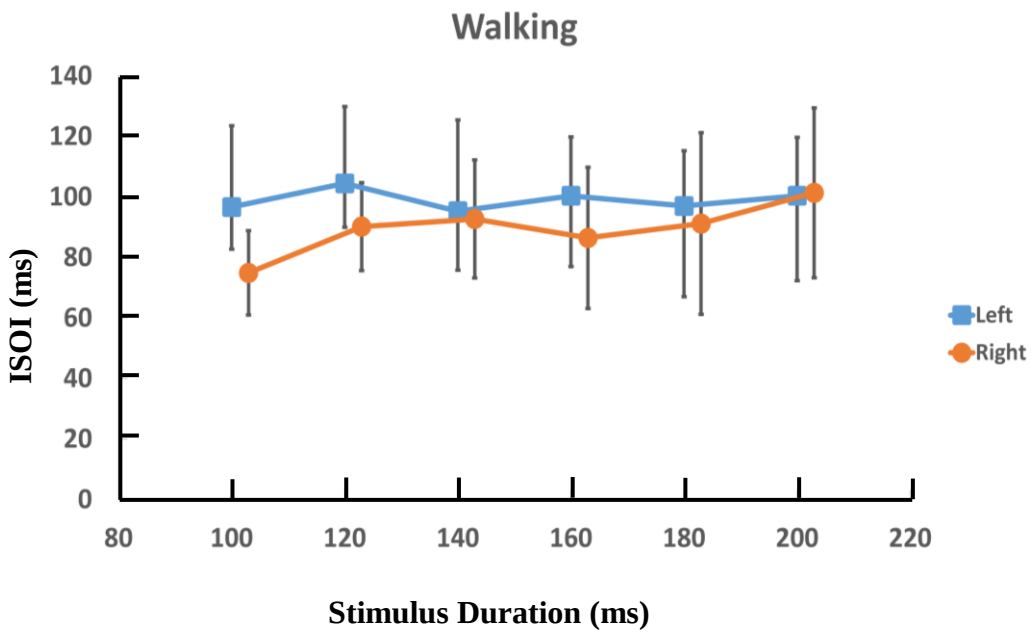

Figure 5.5. The mean and 95% confidence intervals for interstimulus onset interval (ISOI) values giving the "best apparent movement" during walking at difference stimulus durations for left and right directional cues.

A two way repeated measures ANOVA looking at the stimulus duration and activity (standing and walking) as factors showed that stimulus duration had a significant main effect on the mean ISOI, $F(5,95) = 4.448$, $p < 0.05$. Post hoc test using Bonferroni correction showed

Table 5.1: Figure showing the six optimal stimuli combinations used during standing and walking.

Stimuli	SD (ms)	Mean ISOI (ms)	
		Standing	Walking
1	100	64	86
2	120	80	97
3	140	89	94
4	160	89	93
5	180	97	94
6	200	113	101

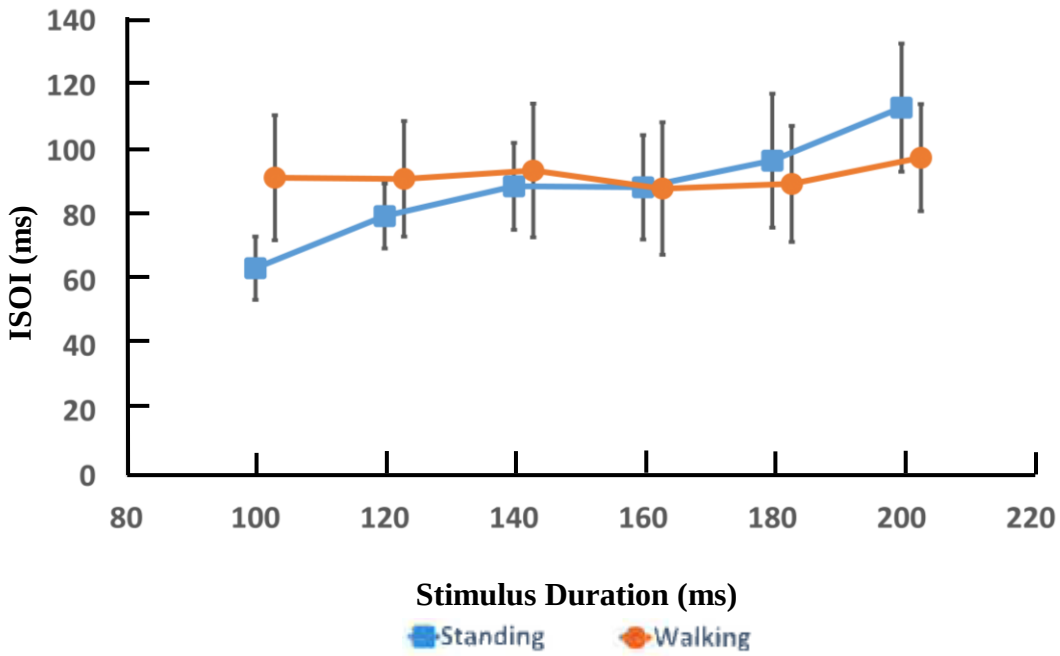

Figure 5.6. Combined mean and 95% confidence intervals for interstimulus onset interval (ISOI) values disregarding any direction bias.

that the difference was apparent when comparing stimulus durations of 100 ms and 200 ms, $p < 0.05$. Activity by itself did not have a significant effect on the mean ISOI, $F(1, 19) = 0.180$, $p > 0.05$. There was an interaction between the stimulus duration and the activity, $F(5, 95) = 3.145$, $p < 0.05$. Exploring this interaction, a one way repeated measures ANOVA with stimulus duration as the factor of interest for the standing experiments showed that stimulus duration had a significant effect on the ISOI, $F(3.523, 66.931) = 8.159$, $p < 0.05$. Evaluating the walking experiments using the same method, the authors observed that stimulus duration did not significantly change the ISOI, $F(5, 95) = 0.275$, $p > 0.05$.

5.4.2 Experiment 2

Friedman tests were performed on the accuracy of the six different stimuli combinations during standing and walking (Figure 1.1). No statistically significant difference was detected during standing $\chi2(5) = 2.79$, $p = 0.733$ ($p > 0.05$). A statistically significant difference was detected during walking $\chi2(5) = 12.9$, $p = 0.0241$. Pairwise comparisons using Conover's test with a Bonferroni correction revealed that the differences were significant between stimulus 1 vs 5 ($p = 0.0012$) and stimulus 1 vs 6 ($p = 0.0165$).

Performing Exact Wilcoxon-Pratt signed-rank tests on the six stimuli pairs between standing and walking (Stimuli 1 – standing and stimuli 1 – walking etc.) did not yield any significant differences ($p > 0.05$).

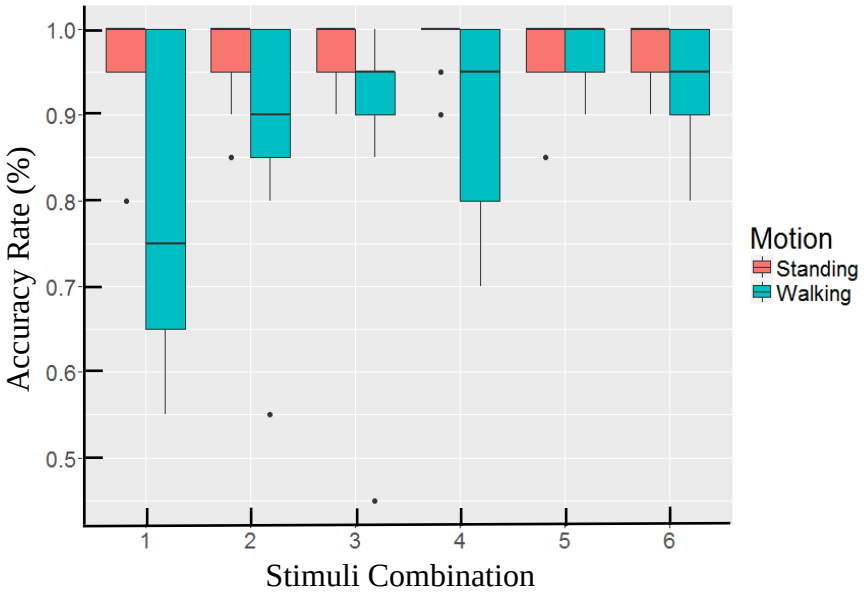

Figure 5.7. Boxplots showing the median accuracy rates, 25 and 50% quartiles, lower and upper whiskers representing the lower and upper quartiles ± 1.5 x interquartile range during standing and walking.

Friedman tests were also performed on the response times of the participants for the six different stimuli combinations during standing and walking (Figure 1.1). No significant differences were detected for both cases of standing and walking, $\chi2(5) = 8.17$, $p = 0.147$ ($p > 0.05$) and $\chi2(5) = 10.8$, $p = 0.0546$ ($p > 0.05$) respectively. Although the difference during walking was not statistically significant, the p-value just above α of 0.05 suggests that a significant difference may be detected if the experiment had a larger sample size.

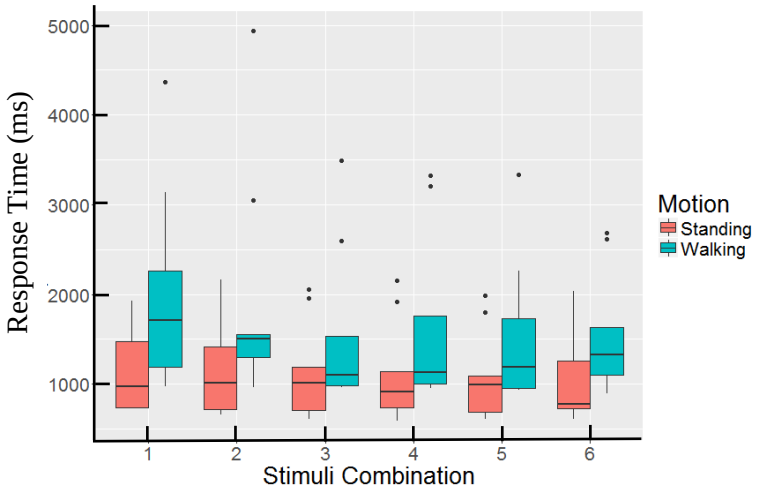

Figure 5.8. Boxplots showing the median response times, 25 and 50% quartiles, lower and upper whiskers representing the lower and upper quartiles ± 1.5 x interquartile range for standing and walking.

Performing Exact Wilcoxon-Pratt signed-rank tests on the six stimuli pairs between standing and walking (Stimuli 1- standing and stimuli 1 – walking etc.) yielded statistically significant differences for pairs 1 and 2 ($p < 0.05$) as seen in Table 1.1.

5.5 Discussion and Conclusion

In this study, we performed two experiments to explore the feasibility of using tactile apparent movement as a feedback method for the lower limb during walking. The first experiment allowed us to quantify the ISOI and stimulus duration combinations, which rendered good perceptions of apparent movement. In general, a longer stimulus duration gave rise to a higher ISOI, which is consistent with previous research by Kirman, Sherrick and Rogers [91, 93]. Likewise, results from the walking experiments were compiled, although increasing the stimulus duration did not seem to lengthen the mean ISOI. The second experiment evaluated the six combinations obtained from experiment 1, by measuring the directional perception accuracy and response time.

During standing, any combination from experiment 1 would deliver similar high median accuracy rates (100%) and low response times (< 1010 ms). During walking, however, combinations with higher stimulus durations yielded higher accuracies and lower response times. From these results it is clear that during walking, the perception accuracy and response times degrades compared with the standing case. The reason for this is currently unclear but could be due to increased cognitive load (needing to interpret sensations while moving),

secondary vibrations from the experimental treadmill or tactile suppression, whereby tactile thresholds increase during movement.

We intend to design a compact ankle bracelet which is capable of administering tactile apparent movement and evaluate the ability for individuals to adjust certain gait parameters based on the feedback they receive.

Table 5.2: Results of the paired Exact Wilcoxon-Pratt signed-rank tests, with * denoting the pairs (stimulus 1 standing and stimulus 1 walking etc) which were significantly different.

Tested Pairs (Standing vs Walking)	Z	P - value
1	-2.07	0.0391*
2	-2.07	0.0391*
3	-1.84	0.0742
4	-1.84	0.0742
5	-1.84	0.0742
6	-1.95	0.0547

Chapter 6: Prototype of a Multi-modal Haptic Feedback Device

In designing a wearable multi-modal haptic feedback device, lateral skin stretch (Chapter 4) and TAM (Chapter 5) were explored as potential modalities to combine with binary vibrotactile feedback for retraining two gait parameters on the lower limb. Both forms of feedback were able to yield very high accuracy rates of up to 100% when the participant was static. TAM was also experimented on individuals during walking, where optimal combinations of SD and ISOI gave high accuracy rates of up to 100% as well. Despite lateral skin stretch being very effective in static settings with a grounded experimental set up, difficulties arise when implementing this feedback in a wearable device.

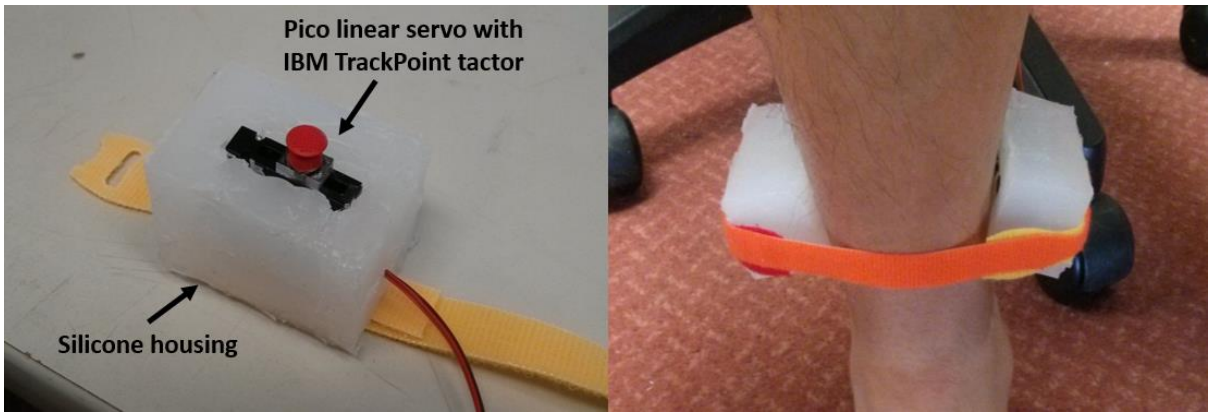

Figure 6.1. Pico linear servo embedded inside silicone housing (left). First skin stretch prototype strapped to leg (right).

An initial device utilising lateral skin stretch was prototyped (Figure 1.1). However, pilot tests revealed that its effectiveness was limited by Newton's 3rd law of motion; every force has an equal and opposite force. One major difficulty of transferring the haptic modality of skin stretch from a grounded set up to a wearable device is that there is an opposing force which can cause masking to occur, as illustrated in Figure 1.1. Masking is when the presence of another stimulus degrades the perception of the intended stimulus [187]. Two ways in which the opposing stretch can be minimised are increasing the mass of the housing and the other is making the bracelet tighter (increasing the normal force against the skin). Although these are potential changes that could be made to minimise masking, increasing the mass is counter-productive for a wearable haptic device and likewise increasing the normal force against the skin. Requirements and specifications to consider for haptic devices as suggested in the literature include the device being lightweight and fitting under clothing [64, 188-191]. These requirements, however, would be negatively affected by an increase in mass of the device if

that indeed was the design decision made to reduce masking. Likewise, increasing the normal force against the skin was experienced by pilot users as being uncomfortable and could also lead to sensory adaptation and thus reducing perception.

Integrating TAM with binary vibrotactile feedback into a single device is advantageous as both stimuli can be elicited using the same actuator. By using the medial and lateral ERM motors for binary vibrotactile feedback and all four ERM motors for TAM, it is possible to eliminate problems of spatial and temporal masking. The use of ERM motors also means the device is mechanically and electrically simple compared with linear actuators (a requirement for lateral skin stretch) [192], and thus minimizing the complexity involved and reducing the size of the overall device. The prototype is introduced in Chapter 4, and is evaluated through the application of a gait retraining task modifying step width and foot progression angle.

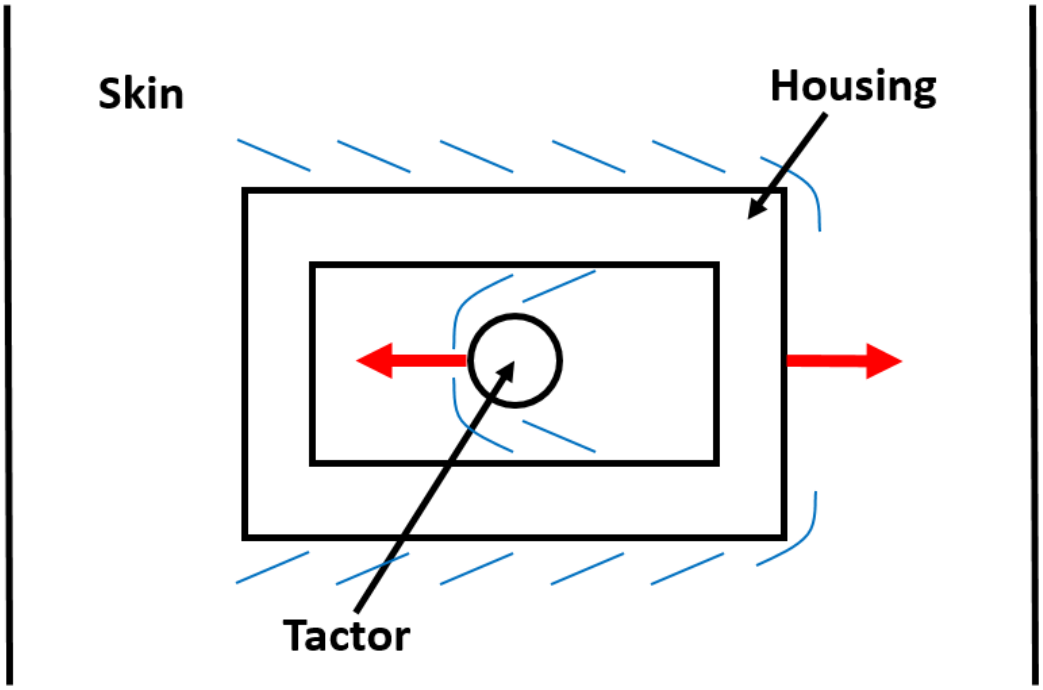

Figure 6.2: Skin stretch device showing linear stretching force of the tactor but also the opposing force of the housing. Local strains are illustrated with blue lines showing potential masking.

Chapter 7: Wearable Lower Limb Haptic Feedback Device for Retraining Foot Progression Angle and Step Width

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Abstract - Technological developments in the last decade have enabled the integration of sensors and actuators into wearable devices for gait interventions to slow the progression of knee osteoarthritis. Wearable haptic gait retraining is one area which has seen promising results for informing modifications of gait parameters for reducing knee adduction moments (KAM) during walking. Two gait parameters which can be easily adjusted to influence KAM include foot progression angle (FPA) and step width (SW). The purpose of this study was to: (1) determine whether a haptic ankle bracelet using binary vibrotactile or tactile apparent movement feedback, could guide individuals to walk with a modified FPA and SW within a short training session (< 15mins); and (2) whether there was a difference between binary vibrotactile feedback or tactile apparent motion feedback for retraining the two parameters. Retraining multiple gait parameters using a single device was a novel aspect of this work and we found that nine out of ten participants were able to retrain their gait using the ankle bracelet in both feedback schemes to within 2° and 39 mm of the target FPA and SW, respectively. We also found no difference in the learning time between the two schemes. Future research will investigate the device performance with patients with knee osteoarthritis and the effective change in KAM by modifying a combination of FPA and SW.

7.1 Introduction

In the last few decades, gait retraining has been used in rehabilitation as a non-invasive intervention for people with abnormal gait either from post-stroke, cerebral palsy, or from hip

and knee osteoarthritis [60]. With traditional gait retraining being confined to clinics and motion capture laboratories, wearable sensing and feedback devices have been an area of interest with the motivation of being able to treat gait abnormalities outside these settings. A review by Shull et al. [155] surveyed current wearable feedback methodologies for abnormal gait interventions. The three methods presently used include haptic (touch) [4, 6], auditory [57, 58], and visual feedback [7, 8]. Although auditory and visual feedback have been successfully implemented, the advantages of haptic feedback is apparent as it allows for the user to free up their auditory and visual sensory channels for other tasks.

We are particularly interested in wearable haptic gait retraining for slowing the progression of knee osteoarthritis (OA). Current gait modifications seek to redirect loads away from the medial tibiofemoral compartment as this is where the disease most commonly occurs [16]. The knee adduction moment (KAM) has been a surrogate measure of knee joint forces because joint forces cannot be measured directly. Thus, reducing the KAM in knee OA patients by modifying gait parameters is one of the intended strategies of slowing the progression of this disease via gait retraining [193, 194].

Previous gait retraining studies for knee OA modified kinematic parameters such as foot progression angle (FPA) [136], tibia angle [4], trunk sway [4], and medial shift of the centre of pressure [5]. It has been suggested through a computational model for level walking [128], stair ascent [141] and descent experiments [139, 140] that step width (SW) is also an influential parameter for reducing KAM. Modifying multiple parameters can further reduce the KAM but previous research implementing this strategy required separate devices attached to different joints and segments to achieve this [4]. We therefore sought to implement a single, wearable device, capable of training multiple parameters. Two parameters which we wanted to explore as suitable candidates for this device were FPA and SW, as these parameters can be modified intuitively with a single ankle device. A recent study by Favre et al. [143] with healthy participants found that toeing-in by an average of 11.1°, the first KAM peak decreased and the second peak remained unchanged. Likewise, SW increases of 0.070 m were enough to significantly reduce both the first and second KAM peaks. Shull et al. [136] found that in patients with knee OA, a toe-in gait modification of only 5° was sufficient to reduce the first KAM peak by an average of 13%.

One challenge for informing multiple parameters is the need for additional feedback gestures, which can be easily learnt and perceived during walking. This challenge is further increased when the gestures are located at the same location. Previous feedback given to the lower limb include binary on-off vibrations, which were used on the foot and knee to inform the adjustments of foot progression and tibial angle during walking [4, 181], and saltation on the foot with limited success [181]. Previously, the authors have explored lower limb haptic perception using lateral skin stretch as a modality in a static setting [180] and also the use of the haptic illusion tactile apparent movement (TAM) during standing and walking [195]. TAM is an illusory sensation felt when multiple discrete loci are sequentially stimulated eliciting a stroking feeling [90], and is a promising modality as it can be easily combined with binary vibratory feedback in the same device by increasing the number of vibration motors. TAM has also been shown to achieve high accuracy rates and low response times in a four-actuator array setup when optimal stimulus duration and interstimulus onset intervals were used [195].

The two research questions we addressed were:

1. Can our wearable ankle haptic bracelet that provides simultaneous binary vibrotactile feedback and tactile apparent movement retrain the foot progression angle and step width of healthy participants with a step accuracy of 80% during walking?
2. Is the number of steps required for retraining different depending on which feedback modality was used for foot progression angle and step width?

7.2 Methods

This study sought to determine whether or not participants could be trained to adopt a new FPA and SW, how many steps were required to achieve adoption, and whether or not there was a difference between which type of feedback was used for training each parameter (binary vibration for angle, apparent movement for step width and vice versa). The experiments in this study sought to replicate aspects from a similar gait retraining study from Lurie et al. [181], which was reflected in the experimental methodology.

7.2.1 Experiments Setup

The custom-made ankle bracelet (Figure 1.1) which the individuals wore uses an Arduino MICRO development board to control four generic eccentric rotating mass motors (8 mm by

2.5 mm) to illicit four different gestures of left/right binary vibrotactile and left/right tactile apparent movement. The ankle bracelet was worn by the individuals on the right leg, with the actuators adhered to the skin using double sided tape and placed horizontally evenly distributed along the anterior hemisphere of the leg at a height halfway between bottom of the heel, and the edge of the calcaneal tendon, as in our previous study [195]. The electronics were held in a neoprene bracelet which was strapped above the motors, seen in Figure 1.1. When binary vibrotactile feedback was given, a double pulse of 250 ms with a 50 ms gap between was given the participants via the left or right motors. In our pilot tests, individuals found it easier to perceive a double pulse as opposed to a single continuous vibration. During left and right tactile apparent movement, the stimulus duration and interstimulus onset interval used was 200 ms and 100 ms, respectively, which we previously found to provide optimal feedback while keeping the total stimulation duration under 1s [195]. A pull feedback modality (move towards the stimulus) was used in this experiment for both binary vibration and TAM feedback as this had been shown in previous studies to be effective [181]. A Vicon motion capture system

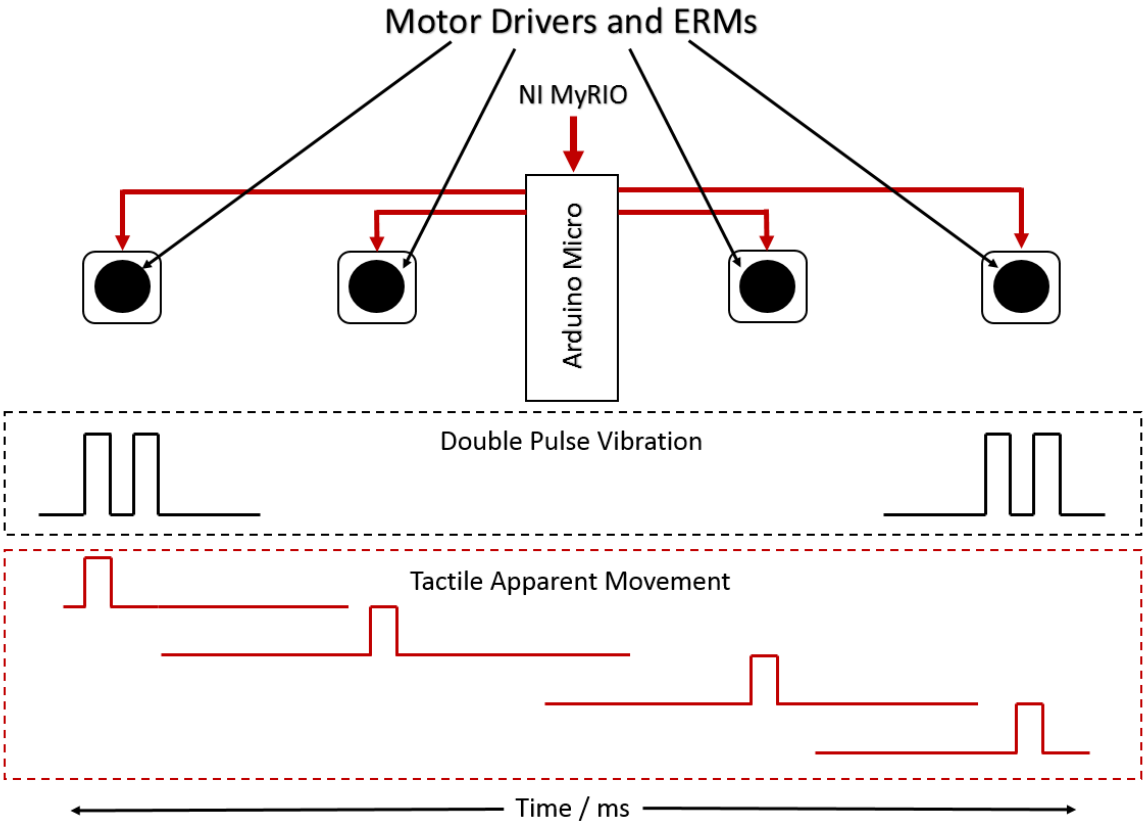

Figure 7.1. Functional block diagram of haptic ankle bracelet. Arduino micro serves as the microcontroller driving the four eccentric rotating mass motors (ERMs). A National Instruments (NI) MyRIO DAQ card sends the control signals from LabVIEW to the Arduino micro. The two feedback modalities of binary vibrotactile and tactile apparent movement are shown. Square waves in diagram are not to scale.

(Vicon Motion Systems, UK) operating at 100 Hz, sampled the marker data into a PC via LabVIEW where the data were processed and control signals sent through a National Instruments MyRIO 1900 DAQ to the Arduino MICRO in the ankle bracelet. The experiments were all conducted on a Force-Instrumented Treadmill (Bertec Corp, MA, USA), as illustrated in the block diagram in Figure 1.1. A treadmill speed of 1.2 m/s was chosen as this represented a comfortable walking speed.

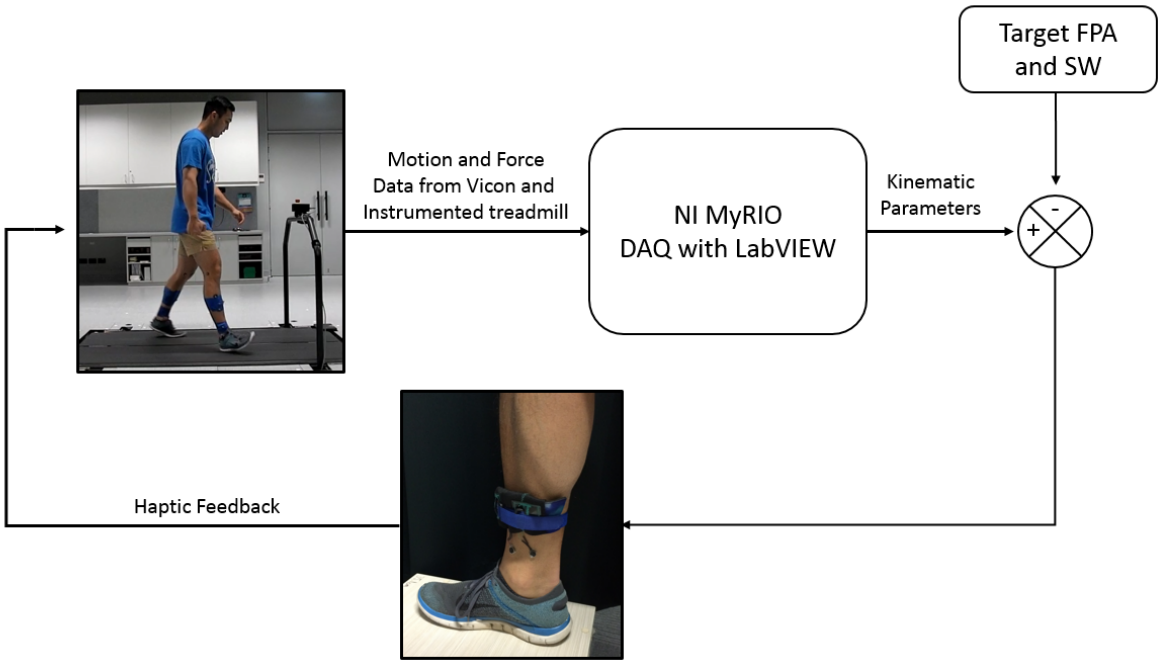

Figure 7.2. Block diagram of full real-time gait retraining system. The PC acquires the marker and force data from the Vicon and Bertec treadmill via the NI MyRIO DAQ. The kinematic parameters are calculated from the raw data acquired, which are compared with the target parameters. Control signals are sent from the NI MyRIO DAQ to the microcontroller of the bracelet as determined by the differential between target and measured FPA and SW.

7.2.2 Experimental Procedure

Ethics committee approval was obtained and ten healthy individuals aged 23 to 43 years gave written consent to participate in this experiment. Three of the individuals had previously been exposed to the haptic bracelet. Motion capture markers were placed on the calcaneus and the second distal phalanx on both the individuals' shoes. Additional markers were placed on the lateral side of the shoes to construct a model within Vicon minimise marker occlusion. FPA was calculated by using the calcaneus to toe vector with respect to the global coordinate frame as reference. Zero degrees was defined as being straight ahead, positive angles when the foot was pointing toe-out and negative when the foot was pointing toe-in. SW was calculated as the width between the calcaneus markers for the left and right foot. Both FPA and SW were

calculated during double support of stance phase specifically foot flat of the right foot. A crossover design was employed, with half the individuals performing feedback scheme 1 first, and then feedback scheme 2 second, whereas the other half of the individuals performed the experiments in reverse order. A washout period of at least two days was given for each participant between the two feedback schemes. Feedback scheme 1 was defined as giving binary to FPA and apparent movement to SW, with scheme 2 being the reverse. A single double pulse for binary feedback and a single stroke from TAM was given during each stance phase if the feedback was outside the target tolerance. A priority feedback scheme shown in Figure 1.2 was implemented as suggested by Lurie et al. [181]. This prioritises FPA as the parameter to be trained first before feedback is given for SW. If at any point in time, the participant steps outside the target FPA tolerance then the system reverts back to training the FPA before SW. Changes in one parameter often leads to involuntary secondary changes in another parameter. FPA was chosen to be the first parameter in the priority scheme due to it having a greater change on SW than the reverse; this has also been found in a previous study by Favre et al. [143]. A correct step is acknowledged when both FPA and SW are within the target tolerance. The accuracy was calculated in real-time of the last 20 steps of the right foot, and completion of the gait retraining was set at achieving 80% accuracy. If accuracy of the participant was not improving and the step count exceeded 500 steps, the trial was ended and deemed unsuccessful. All the feedback modalities were presented to the individuals before the experiment and the interpretation of the gestures was clearly shown. Individuals were not given information on what strategies to use in modifying their gait but were given the feedback sensations beforehand and were told which gait parameter it was associated with. Individuals practiced making gait modifications while standing until the authors were satisfied the individuals understood what was required of them.

Before the experiment, individuals were told to walk with normal gait on the treadmill for 50 steps. During this period, a baseline average of their FPA and SW was calculated from their last 20 steps. The targets used for the experiment was thus baseline FPA + 5° and SW + 85 mm. These targets were deemed reasonable without causing too much discomfort but enough to be outside the normal walk gait parameters taking into account step variance. FPA targets had a tolerance of ±2°, which was similar to a previous observation in the literature about the foot angle variation during walking [196]. Owings et al. [197] found that the SW variability of older adults was 2.5 ± 0.7 cm (mean ± standard deviation). The tolerance for SW used in our study of 39 mm was therefore the upper end of the 95% confidence interval of SW

variability of older adults which the author's felt conservatively encompassed all the individuals. The root mean square error (RMSE) was also determined for FPA and SW for each trial. Exact Wilcoxon-Pratt Signed-rank tests were performed ($\alpha = 0.05$) to determine whether or not there were differences in the feedback schemes and RMSE of FPA and SW. RMSE for FPA and SW was calculated as follows, noting that the RMSE for SW was only calculated if FPA was outside the target tolerance:

$$RMSE(FPA) = \sqrt{\frac{\sum_{i=1}^{n}(actual\ FPA_i - target\ FPA_i)^2}{n}}$$

$$RMSE(SW) = \sqrt{\frac{\sum_{i=1}^{n}(actual\ SW_i - target\ SW_i)^2}{n}}$$

i = step count, n = total number of steps during a single trial

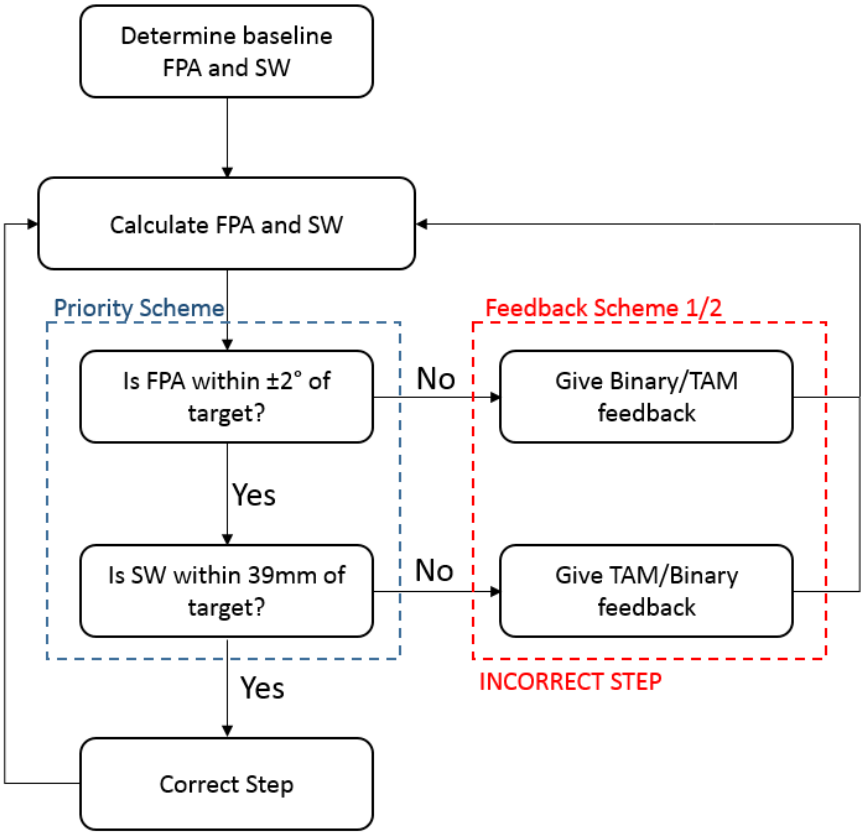

Figure 7.3: Flow chart detailing the priority feedback scheme used. Feedback Scheme 1 is giving binary vibrotactile feedback to FPA and tactile apparent movement (TAM) to SW, whereas feedback Scheme 2 is the reverse. Both requirements of being within the target FPA and SW are required for the step to be deemed as 'correct', otherwise feedback is given and the step is deemed 'incorrect'.

7.3 Results

All individuals successfully completed the trial for feedback scheme 1 and nine of the ten individuals completed the trial for feedback scheme 2 as shown in Table 1.1. There was no difference in the number of steps to complete the trial between the two feedback schemes when we excluded participant 10 with the incomplete trial (Z = -0.178, p = 0.910), Figure 1.1. When we included participant 10 by replacing the 'X' (Table 1.1) with a dummy step number of 1000 to ensure it had maximum rank in our statistical test, the same conclusion can be drawn, (Z = 0.357, p = 0.770). The root-mean-squared-error (RMSE) of FPA and SW were also calculated (excluding participant 10) and no statistically significant difference was found between the RMSE of the two feedback schemes for both FPA (Z = 1.2472, p = 0.2344) and SW (Z =

Table 7.1. Tables showing the normal (mean ± SD) and target (target ± tolerance) gaits of each participant during both feedback schemes along with the number of steps it took for completion.

A: Feedback Scheme 1

Subject	Normal Gait		Target Gait		RMSE of FPA	RMSE of SW	Completion Steps
	FPA / °	SW / mm	FPA / °	SW / mm			
1	6.6 ± 3.0	126 ± 20	11.6 ± 2.0	211 ± 39	2.5	26	142
2	8.9 ± 6.4	105 ± 13	13.9 ± 2.0	190 ± 39	4.6	35	133
3	7.0 ± 1.9	161 ± 14	12.0 ± 2.0	246 ± 39	7.1	24	263
4	8.5 ± 1.3	112 ± 11	13.5 ± 2.0	197 ± 39	3.3	23	103
5	4.1 ± 1.6	158 ± 15	9.1 ± 2.0	243 ± 39	9.8	41	121
6	6.7 ± 1.4	137 ± 13	11.7 ± 2.0	222 ± 39	3.1	50	393
7	13.3 ± 1.8	104 ± 12	18.3 ± 2.0	189 ± 39	2.7	19	43
8	8.2 ± 3.1	107 ± 12	13.2 ± 2.0	192 ± 39	2.3	39	189
9	7.1 ± 1.7	161 ± 21	12.1 ± 2.0	246 ± 39	3.2	30	181
10	4.4 ± 3.8	158 ± 27	9.4 ± 2.0	223 ± 39	3.2	11	276

B: Feedback Scheme 2

Subject	Normal Gait		Target Gait		RMSE of FPA	RMSE of SW	Completion Steps
	FPA / °	SW / mm	FPA / °	SW / mm			
1	4.5 ± 1.5	163 ± 12	9.5 ± 2.0	248 ± 39	2.7	29	122
2	11.0 ± 3.0	104 ± 23	16.0 ± 2.0	189 ± 39	5.2	27	74
3	5.8 ± 4.8	126 ± 12	10.8 ± 2.0	211 ± 39	4.6	17	173
4	9.3 ± 3.8	69 ± 14	14.3 ± 2.0	154 ± 39	4.1	32	504
5	1.9 ± 1.6	142 ± 16	6.9 ± 2.0	227 ± 39	3.5	24	125
6	3.9 ± 3.2	156 ± 19	8.9 ± 2.0	221 ± 39	3.8	40	282
7	14.6 ± 2.4	110 ± 14	19.6 ± 2.0	195 ± 39	4.8	30	66
8	6.4 ± 1.2	152 ± 15	11.4 ± 2.0	217 ± 39	8.1	40	135
9	3.6 ± 1.5	171 ± 23	8.6 ± 2.0	226 ± 39	5.2	47	802
10	4.1 ± 4.0	169 ± 25	9.1 ± 2.0	254 ± 39	X	X	X

X = Subject did not complete retraining exercise.
Normal gait values were taken was from the subject's last 20 steps.

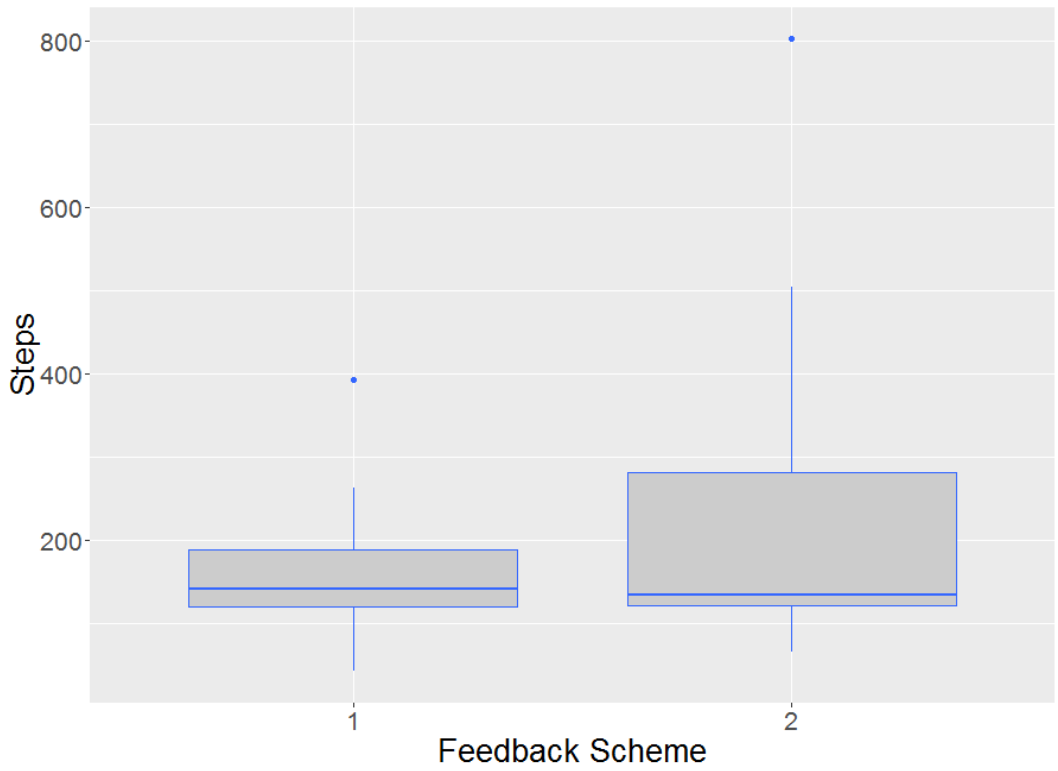

Figure 7.4. Boxplots showing the median completion steps, 25 and 50% quartiles, lower and upper whiskers representing the lower and upper quartiles ± 1.5 x interquartile range for the two feedback schemes.

0.59399, p = 0.9766). Likewise, when participant 10 is included by using 100 as the dummy RMSE for both FPA and SW (to ensure maximum rank), no difference was found for either the FPA (Z = 1.1722, p = 0.275) or SW (Z = 0.45868, p = 0.695). Figure 1.1 illustrates an example of the data collected from a trial.

7.4 Discussion

This study explored the feasibility of retraining walking gait to match FPA and SW using a wearable haptic ankle bracelet. The two feedback schemes investigated in this study were compared with regard to altering foot progression and step width. It was shown that nine out of ten individuals were able to retrain their FPA and SW using our ankle bracelet. As a whole, there was no difference in the feedback schemes in terms of the number of steps required for the individuals to complete the gait retraining exercises ($p > 0.05$).

The one participant who could only complete the trial using feedback scheme 1 mentioned that it was not an issue of perceiving the gestures but more a difficulty with keeping within the ±2° tolerance band of FPA. This participant had performed feedback scheme 2 first.

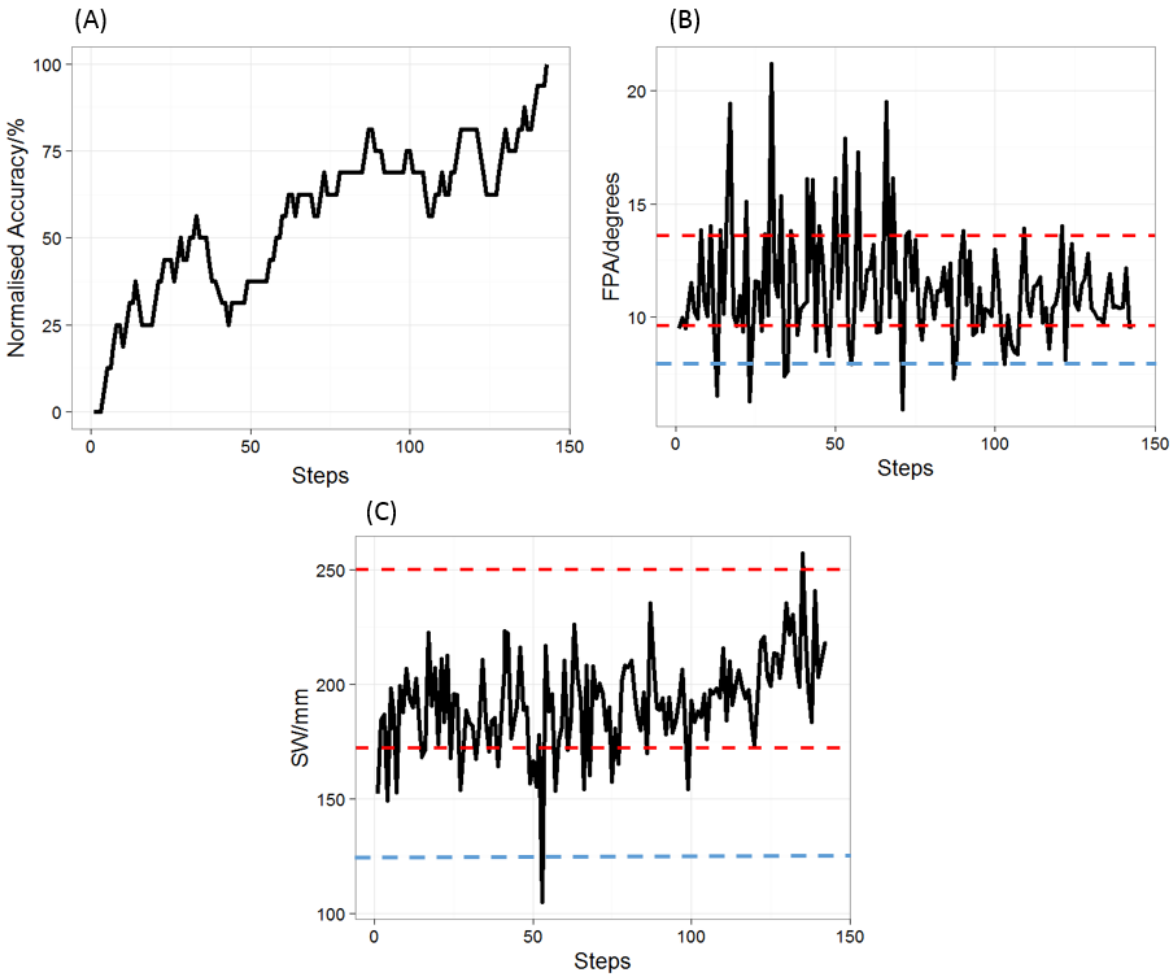

Figure 7.5. (A) Accuracy, (B) FPA, and (C) SW of example subject over duration of trial. Red dash lines show maximum and minimum values considered 'correct' for each parameter. Blue dashed lines show baseline FPA (6.6°) and SW (126 mm) values as calculated from the last 20 steps during the normal walking phase for this participant.

Therefore, it is possible that he learned to adjust the FPA differently when the trial was performed with feedback scheme 1. Shull et al. [136] noted that it was possible to alter FPA by using either the toes or the heel as the pivot. As participants in this study were not told specific strategies or methods to adjust their gait, this may have also been a contributing factor for individuals completing the retraining tasks. Although the ±2° tolerance for FPA was used in a prior study by Shull et al. [136], it is possible that certain individuals may have higher gait variability that increases the difficulty they find the proposed gait retraining experiments.

We are not aware of previous gait retraining studies that have used a single haptic device that combines binary vibration and tactile apparent movement for multi-parameter gait retraining. As such, comparison with the literature is difficult. Lurie et al. explored the use of apparent movement (although incorrectly presented as saltation based on the original definition

[83]) on the back for trunk sway, lateral and medial of the knee and foot for tibial angle and FPA, respectively [181]. The motor array, however, was placed vertically and not horizontally as in this study. They therefore used apparent movement as a binary gesture requiring arrays on either side of the body segment of interest. Apparent movement was not adopted in Lurie's study as they felt it was less perceivable during walking. However, we have previously showed that by using optimal values tuned for walking, there is no difference in accuracy perception between apparent movement and binary vibration [195]. Although the feedback schemes in the current study did not display any difference in terms of the number of steps required for successful gait retraining, Figure 1.1 shows that feedback scheme 1 had less step variance, suggesting it may be more suitable, especially for those participants who took significantly longer in scheme 2. At this point, it is unknown whether certain participants are sensitive to the feedback scheme used. Another experiment with those participants whose steps differed significantly between the two schemes would be required with multiple trials of each scheme to determine whether there really is a difference for those individuals. In the previous study by Lurie et al. [181] individuals were given a total of 300 steps to learn a particular gait and reported that >97% of learning trials were successful when targeting each parameter separately. Targets for each gait parameter and their tolerances were not specified so we are unable to make a direct comparison with our haptic device. However, 90% of our trials were successfully completed in less than 300 steps using feedback scheme 1 in our study. All trials were completed in less than 400 steps, noting that our trials required a successful target range for *both* parameters. Wheeler et al. found that using haptic feedback was equally effective when being compared with visual feedback in a study where individuals were able to freely modify their gait in order to reduce the peak KAM presented to them [6]. A direct comparison of learning rate using visual feedback and our wearable ankle bracelet has not been performed, but would be worth pursuing.

The lack of difference found in the RMSE of FPA and SW during the trials between feedback schemes 1 and 2 shows that in general, individuals did not overshoot the target ranges differently based on the schemes used. One interesting research question would be exploring how different strategies of reaching gait targets affect RMSE and learning rate. If significantly different RMSE's were detected, it could signal that certain feedback schemes and or strategies took longer for individuals to perceive and thus correct. Although this study did not explicitly explore participant gait modification strategy, we observed differences in the way they attempted to change and maintain their gait. Some individuals used methods like visually

locating their foot placement with respect to the treadmill, walking with longer strides, and walking with less knee flexion (stiffer legs). Further experimentation would be needed to understand the strategies used, but these different strategies might explain the wide variability in steps required for completion between individuals.

One limitation in this study is that only healthy individuals under 50 years of age participated in this study. It is possible that patients with knee OA may have difficulty successfully completing the gait retraining tasks set out in this study, although previous research on knee OA patients have shown implementations from healthy individuals to knee OA patients have been successful [157]. Only one target, namely toeing-out by 5° and increasing SW by 85 mm, was used in this study leading to the possibility that certain gait modifications may take shorter or longer to learn. However, the toe-out modifications in this study are consistent in magnitude with previous modifications used in a study with knee OA patients [133]. We could not find suitable magnitudes of SW modification for knee OA patients in the literature. Furthermore, we acknowledge that the completion of the gait retraining task is sensitive to the tolerance band that was used of ±2° for FPA and ±39 mm for SW, where a wider tolerance would have made it easier for the individuals to complete the task. At this point, it is unknown whether these tolerances make it possible for knee OA patients to complete the retraining task. However, individuals who exhibited large gait variations during normal gait did not necessarily take longer to retrain their gait (table 1). Future studies could alter feedback only when the parameter change is less than intended but not when it was more, which was used by Shull et al. [4]. The optimal values for apparent movement determined by the authors in a previous study were also obtained from healthy individuals [195]. Previous studies have shown that vibratory sensitivity decreases with age [198] and also in people with knee OA [199, 200]. It would be appropriate to determine whether optimal values for apparent movement are different for a patient population. It is possible that with a larger sample size, a significant difference between feedback schemes may have been detected, although this would not change the finding that our ankle bracelet was able to successfully alter the FPA and SW in a short training session. Having seen promising results from this study, the authors would like to see how knee OA patients learn to walk with a new gait using this ankle bracelet through a variety of FPA and SW combinations in the future. We expect that less feedback would be required to be given to the individuals with practice using a fading feedback method, and that eventually,

patients will be able to adopt a new gait without the need of feedback which has been observed by Shull et al. [157] in a six week gait retraining study.

Previous studies have prescribed and explored patient specific gait modifications [4, 50, 128]. Therefore, it is likely that the ideal gait modifications and effectiveness of retraining FPA and SW varies individually between individuals. Favre et al. [143] showed that general schemes like increasing trunk sway, increasing SW, and toeing-in can be combined to reduce the KAM, although the required amplitude differs between individuals for a given reduction in KAM. We intend to explore whether the optimal FPA and SW gait for a single individual can be determined using a regression model of the KAM at different FPA and SW along the participant's range of motion. With the long-term goal of implementing real-time wearable gait retraining for patients with knee OA, by optimising the desired gait in the laboratory, all that is required for implementing a real-time wearable haptic feedback system for gait retraining is sensors and algorithms which are able to accurately measure the participant's FPA and SW in real-time, one of which has been proposed for foot progression angle by Huang et al. [55].

Wearable haptic feedback has been demonstrated to be feasible for gait retraining, which can be used outside clinical and laboratory settings [5-8, 10-13]. In summary, this study explored the use of a custom haptic ankle bracelet to retrain FPA and SW in a short training session, and showed that it was effective for retraining two parameters in a single device. Further work to explore the different possible feedback schemes with this device is required.

Chapter 8: Data-driven Method for Reducing Peak Knee Adduction Moments by Altering Foot Progression Angle and Step Width

Abstract – Participant-specific gait modifications have been shown to be effective in reducing peak knee adduction moments (KAMs) which can slow the progression of medial compartmental knee osteoarthritis. Two gait parameters which show promise in reducing peak KAM are foot progression angle (FPA) and step width (SW), but gait recommendations using these parameters and their influences on KAM are currently unknown. The authors use a data-driven correlation method of identifying a subject in which their peak KAM are sensitive to changes in these parameters. Regression models were then created for predicting peak KAMs and the participant was given gait recommendations which reduced their predicted peak KAMs and finally these predictions were compared with actual values from data collected in the motion capture laboratory. In summary, we found a healthy subject who was sensitive to changes in FPA and SW through a short laboratory session of walking on an instrumented treadmill while freely varying their FPA and SW through their range of motion. Regression models adequately provided recommendations to reduce the second KAM peak, but not the first. For the subject of interest, SW was the only parameter which influenced their peak KAM for the range of motion explored.

8.1 Introduction

Knee osteoarthritis (OA) is a common and disabling disease, increasing the difficulty in which activities involving the lower extremity can be performed [5]. The medial compartment of the tibiofemoral joint has the highest occurrence of cartilage lesions leading to the progression of knee OA compared with other regions of the knee joint [16]. A promising, non-invasive intervention to slow the progression of medial compartment knee OA is to redirect loads away from the medial compartment by modifying walking gait.

The knee adduction moment (KAM) is a suitable surrogate measure of the medial compartmental load [1] and has been associated with the onset, progression, and severity of knee OA [2, 3]. Therefore, one strategy of gait retraining for patients with early onset of medial compartmental knee OA is to reduce the KAM during walking [193, 194]. Modifying gait parameters focus on reducing the KAM by decreasing the moment arm of the ground reaction force to the knee centre, which can be achieved through various kinematic adjustments [46].

Previous gait retraining studies have explored modifications to several parameters, including; the centre of pressure of the foot [5], tibial angle [4], foot progression angle [136] and trunk sway [4, 138]. Increased trunk sway can reduce the maximum and first KAM peak [138]. Shifting the tibia medially during the stance phase of gait has been found to reduce both the first and second KAM peak [50]. Changing the FPA by toeing-out has not been as consistent amongst studies. Several studies show that toeing-out decreases the second KAM peak [126, 127] but the first KAM peak has been reported to increase [126, 135], decrease [133, 134], or have no change at all [129]. Likewise, there is no consensus for the influence of toe-in gait on the effect of the first and second KAM peaks. Studies have shown to increase both KAM peaks while toeing-in [134], increase just the second KAM peak but not the first KAM peak [127], decrease the first KAM peak with the second peak unchanged [127, 136], or toe-in having no effect on KAM peaks at all in some individuals [127]. Step width has also been identified as a gait parameter which could reduce KAM [128, 139, 141]. An increase in step width has been shown in patient-specific models to reduce KAM peaks [128, 130], although this gait modification has not been attempted in an experimental study.

The participant-specific variability in the KAM following gait modifications highlights the need for personalized prescriptions, as well as real-time, accurate feedback to achieve the desired modification [4, 50]. To this end, we have developed a haptic ankle bracelet to simultaneously retrain FPA and SW during walking [195]. These two parameters are easily modifiable and have the potential to reduce the KAM, although, it is not known how these parameters interact and influence the KAM. This paper comprises a data-driven approach to identify individuals whose peak KAM may be sensitive to changes in FPA and SW, construct regression models predicting the first and second KAM peaks, and seeing how actual gait changes compare with initial predictions. The two null hypotheses we wanted to test were, H_0 = No difference between calculated KAM peaks and predicted KAM peaks and H_0 = No difference between the calculated KAM peaks of normal vs modified gaits.

8.2 Methods

Experiment 1: Individuals walked on a treadmill making random variations to their foot progression angle and step width. Correlation matrices of gait parameters against KAM peaks were generated to identify candidates who showed FPA and SW having an influence on their KAM peaks. One participant was chosen based on their strong correlations and their

participant-specific regression models were constructed. The prescribed gait modifications for FPA and SW which reduced either the first or second KAM peaks were used in Experiment 2.

Experiment 2: Using the gait recommendations identified from Experiment 1, the same participant walked with new gaits. KAM peaks were calculated while the participant walked at their new gaits to determine whether or not their peaks were significantly reduced, how FPA and SW influenced these changes, and how these compared with predicted values from the regression models in Experiment 1.

8.2.1 Experimental Setup

A Vicon Motion Capture System operating at 100 Hz and a Force-Instrumented Treadmill (Bertec Corp, MA, USA) samping at 1000 Hz captured the motion and ground reaction forces generated by individuals as they walked at a 'comfortable' speed of 1.2 m/s. A National Instruments MyRIO DAQ system and a computer with LabVIEW was used in Experiment 2 for real-time visual feedback, assisting the individuals to walk with their new gait while marker and ground reaction forces were being captured.

Retroreflective markers (9.5 mm) were placed on the calcaneus and the second distal phalanx on both shoes of the individuals. Markers were placed on the medial and lateral epicondyle of the tibia. Markers placed on the medial and lateral malleoli were used to determine the ankle joint centre and tibial axis. Finally, a four-marker tracking cluster was placed on the shank half way between the knee and ankle joints. The position of the femoral and tibial anatomical landmarks was recorded with respect to these tracking clusters during a static trial and then projected during the dynamic trials to reduce marker position error due to soft tissue artefact. The KAM was calculated by determining the moment at the knee joint centre due to the ground reaction force from the foot's centre of pressure in the frontal plane of the tibia (Figure 1.1).

8.2.2 Experimental Procedure

8.2.2.1 Experiment 1

Eleven healthy individuals (9 males and 2 females, mean height 1.73 m (SD = 0.06 m) and mean mass 73 kg (SD = 13 kg)) with ages ranging from 22 to 42 years gave written consent to participate in the study. Approval for the study was given by the university ethics committee. Individuals were given a minute to walk on the treadmill as a warm up, after which they were instructed to vary their foot progression angle and step width freely along their full range of motion while walking on the treadmill. For foot progression angle that meant varying the toe-in and toe-out angles and for step width that meant varying the width between the feet as long as either foot did not cross the centre line of the treadmill. Individuals walked for at least one minute (at least 28 foot strikes from both feet) and their marker and force plate data were collected for post processing.

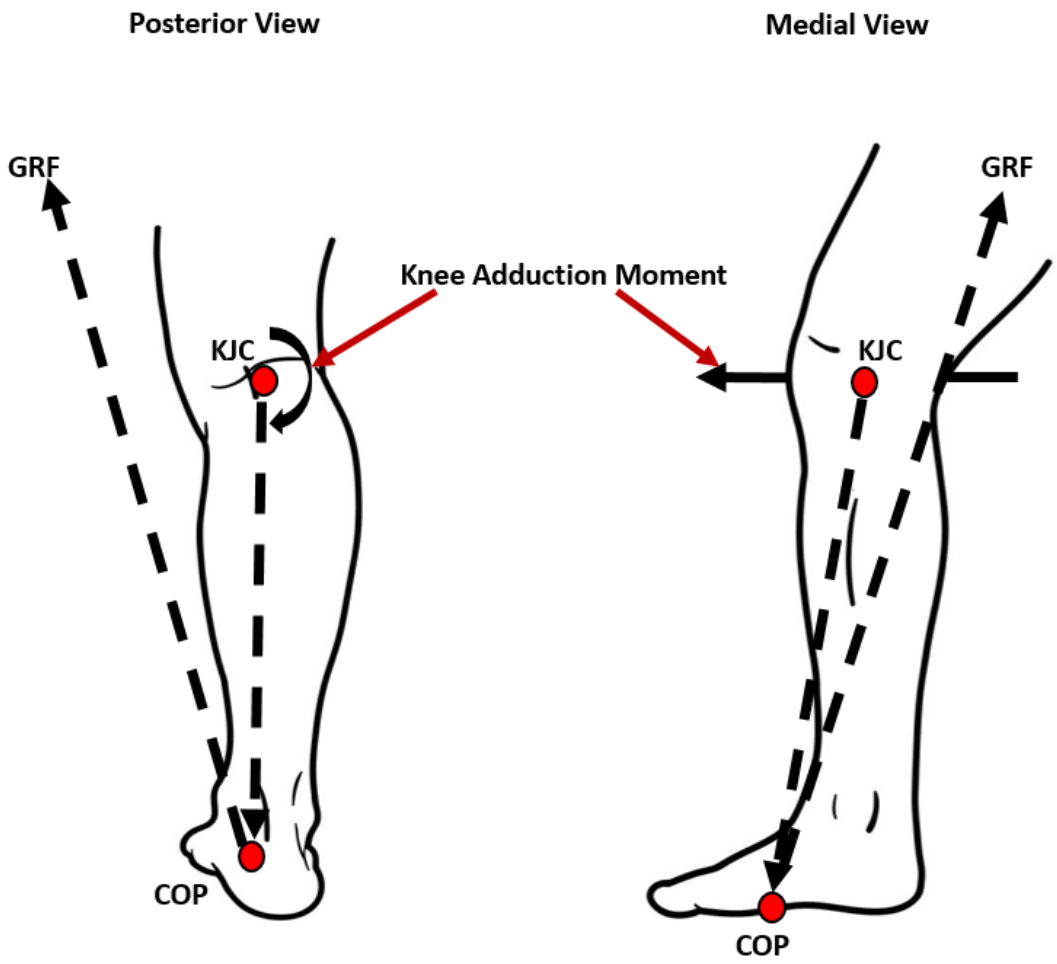

Figure 8.1. Posterior and medial views of vectors involved in the calculating the KAM. Knee Joint Centre (KJC) is defined as the as the midpoint of the medial and lateral epicondyle markers. Centre of Pressure (COP) is the point on the ground where the ground reaction force (GRF) vector is considered to originate.

8.2.2.2 Data Processing and Statistical Analysis

The KAM was calculated for each step, normalized by body mass and height, and the first and second peaks of the KAM during each gait cycle were identified (Figure 1.1). The corresponding FPA and SW of each stance phase were calculated. Scatter matrices were constructed for each leg to determine which participants' legs had a significant change in KAM peaks from both changes to their FPA and SW (an example of which is shown in Figure 1.2). One participant, who had shown strong correlations for both of the peak KAMs with FPA and SW was chosen and regression models for predicting the first and second KAM peak with FPA and SW were constructed. New gait recommendations, which reduced the peak KAMs based off the regression models for the participant (e.g. toe-in and increase step width), were given to the participant in Experiment 2.

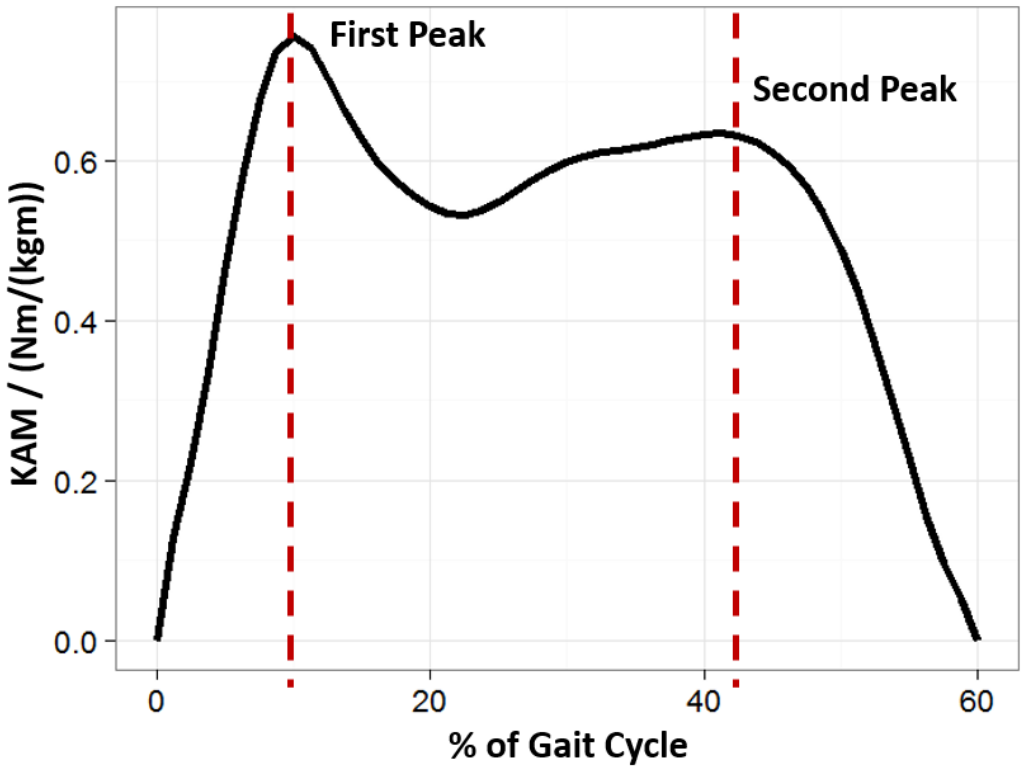

Figure 8.2. Example of a single stance phase showing normalized KAM with respect to % of gait along with the respective first and second KAM peaks.

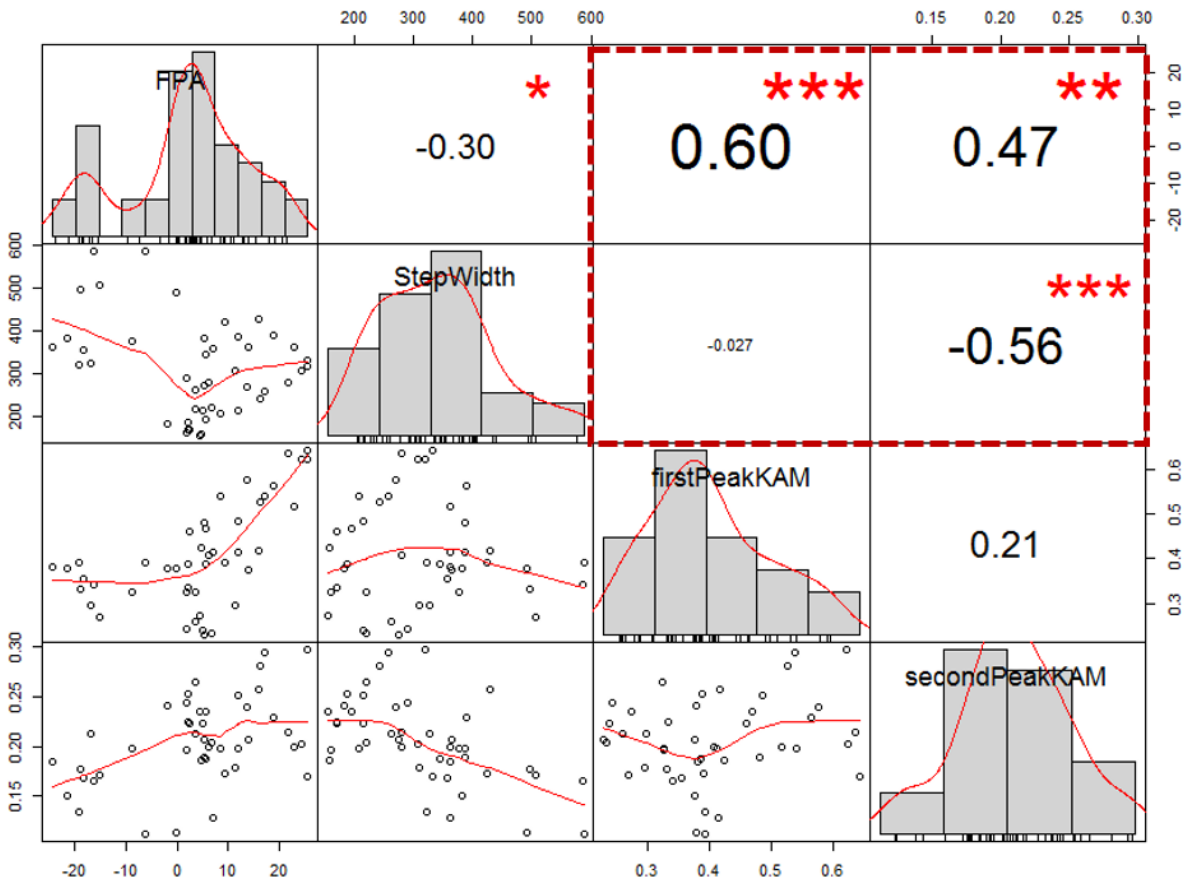

Figure 8.3. Scatter matrix of participant showing linear correlations between FPA, SW, first and second KAM peak. Correlation coefficients are shown in the top right triangle of matrix, with *, **, *** denoting statistical significance at the $p < 0.05$, $p < 0.01$, and $p < 0.001$ levels.

8.2.2.3 Experiment 2

The candidate who showed "promise" for gait retraining was asked to return for a second experiment the next day. The purpose of this experiment was to determine whether the gait recommendations from Experiment 1 would reduce the peak KAM and compare the actual results with our predicted KAM values from the regression models. The participant walked on the treadmill using their normal gait followed by three modified gaits, which were predicted to reduce first and second KAM peaks from Experiment 1. In each modified gait trial, the participant walked on the treadmill for at least two minutes, whereby the mean FPA, SW, and peak KAMs were calculated. Real-time visual feedback showing the target FPA and SW in LabVIEW was given for trials for which it was relevant. The FPA and SW gaits were chosen from the participant-specific regression models as gaits which could reduce the peak KAM but still being gaits which the participant could comfortably adopt.

8.2.2.4 Statistical Analysis

A Kruskal Wallis rank sum with Conover's post hoc test was performed to determine whether there were significant differences between the normal and new gaits. One sample t-tests were used to analyse whether the measured peak means were different from the predicted values for the regression models.

8.3 Results

8.3.1 Experiment 1

Scatter matrices showed scatter plots and linear correlations of foot progression angle and step width against first and second KAM peaks (Figure 1.2). We found large participant-to-participant variation in foot progression angle and step width correlation (Table 1.1). Out of the 22 trials collected (11 individuals with both left and right legs), only 2 out of 22 for the first KAM peaks and 4 out of 22 for second KAM peak had significant ($p < 0.05$) correlations with SW. FPA had a higher proportion of trials with correlations with 12 out of 22 for first and 9 out of 22 for second KAM peaks.

Table 8.1. Proportion of trials (11 individuals, left and right legs) showing statistically significant correlations between gait parameters and peak KAMs.

	First Peak	Second Peak
FPA	$\frac{12}{22}$	$\frac{9}{22}$
Step Width	$\frac{2}{22}$	$\frac{4}{22}$

Five out of the 22 trials had significant correlations between at least one KAM peak against both FPA and SW. One participant from these five was chosen to perform Experiment 2. For this participant, the regression models were generalised additive models which predicted

new gaits for reducing the KAM peak of interest. Regression models predicting the first and second KAM peaks (Figure 1.1) were selected using the Bayesian Information Criterion.

The first KAM peak was not sensitive to modifications of SW as a gait parameter but toeing-in reduced the KAM (Figure 1.1). However, both FPA and SW reduced the second KAM peak, which was reduced when the participant walked with toe-in and a wider SW. Based on the two models created for prescribing a new gait recommendation for the participant, by walking toe-in with a wider SW, the participant was predicted to have a lower first and second KAM peak.

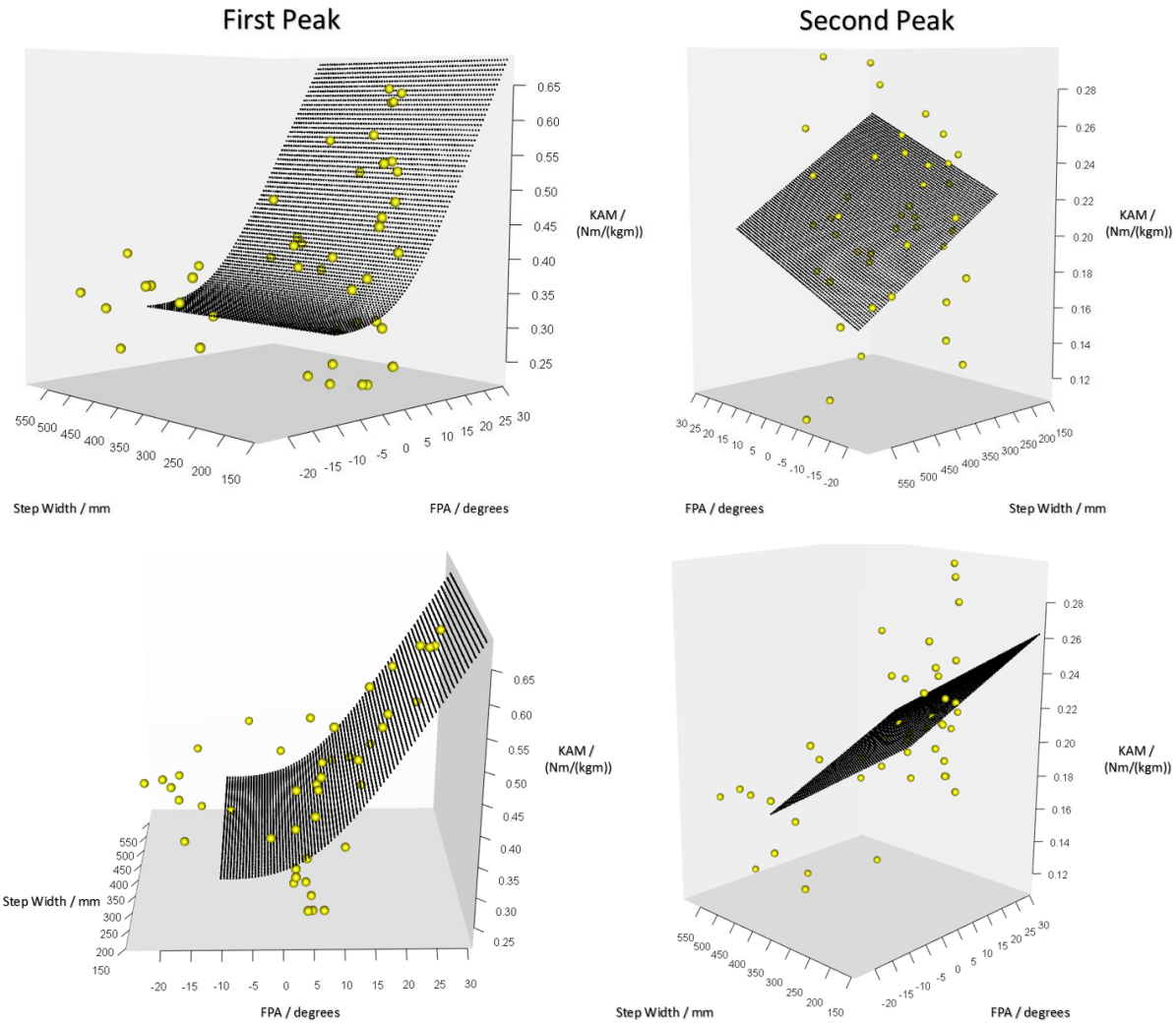

Figure 8.4. Generalized additive regression models predicting the first (left, R^2 = 0.57) and second (right, R^2 = 0.39) peak KAMs selected by the Bayesian Information Criterion at different FPA and SW for the chosen participant.

8.3.2 Experiment 2

Regression models illustrate how the three gait modifications influence the peak KAM values from normal gait (Figure 1.1). One sample t-tests show that all mean KAM peaks were significantly different from the predicted mean values from their respective regression models (Table 1.1). A Kruskal Wallis rank sum test was performed on the first and second peak KAMs for the four different walking gaits, $\chi2(3) = 162.28$, $p < 0.001$ and $\chi2(3) = 128.49$, $p < 0.001$ respectively. Table 1.2 shows the three new gaits compared with the normal gait.

Table 8.2. Predicted and calculated first and second KAM peaks for the different gaits perturbing toe-in and step width (SW) from Figure 1.1. * denotes statistically significant differences ($p < 0.001$) between the predicted and measured values of the corresponding KAM peak. All values shown are mean ± SD.

Gait (FPA / SW)	Actual FPA (°)	Actual SW (mm)	**Predicted** KAM (Nm/(kg x m))		**Measured** KAM (Nm/(kg x m))	
			First Peak	Second Peak	First Peak	Second Peak
Normal (7.4°/165mm)	7.4 ± 1.7	165 ± 41	0.39	0.24	0.43 ± 0.05*	0.28 ± 0.02*
Toe-in with normal SW (2°/165mm)	1.5 ± 1.0	166 ± 12	0.35	0.23	0.43 ± 0.05*	0.29 ± 0.03*
Normal FPA with wide SW (7.4°/500mm)	7.3 ± 1.0	479 ± 11	0.38	0.18	0.57 ± 0.08*	0.23 ± 0.04*
Toe-in with wide SW (2°/500mm)	1.6 ± 1.0	484 ± 15	0.35	0.17	0.59 ± 0.10*	0.24 ± 0.03*

Table 8.3. Actual calculated first and second KAM peaks along with the Conover's post hoc results when compared with normal gait. * denotes statistically significant differences ($p < 0.001$) between corresponding KAM peaks of the new gait as compared with the participant's normal gait. Match pair superscripts (A and B) denote no significant difference ($p > 0.05$) within their respective peak groups. All values shown are mean ± SD.

Gait (FPA / SW)	New Gait				Normal			
	Actual FPA (°)	Actual SW (mm)	KAM (Nm/(kg x m))		FPA (°)	SW (mm)	KAM (Nm/(kg x m))	
			First Peak	Second Peak			First Peak	Second Peak
Toe-in with normal SW (2°/165mm)	1.5 ± 1.0	166 ± 12	0.43 ± 0.05	0.29 ± 0.03	7.4 ± 1.7	165 ± 41	0.43 ± 0.05	0.28 ± 0.02
Normal FPA with wide SW (7.4°/500mm)	7.3 ± 1.0	479 ± 11	0.57 ± 0.08*A	0.23 ± 0.04*B	7.4 ± 1.7	165 ± 41	0.43 ± 0.05	0.28 ± 0.02
Toe-in with wide SW (2°/500mm)	1.6 ± 1.0	484 ± 15	0.59 ± 0.10*A	0.24 ± 0.03*B	7.4 ± 1.7	165 ± 41	0.43 ± 0.05	0.28 ± 0.02

8.4 Discussion

In this study, we used a data-driven method to explore the influence of foot progression angle and step width on peak KAM to identify individuals in which they were effective gait retraining parameters. A participant was selected from the group which showed strong correlations for

both parameters and participant-specific regression models were used to prescribe new gait parameters which were likely to reduce their first and second KAM peaks. Based on the gait

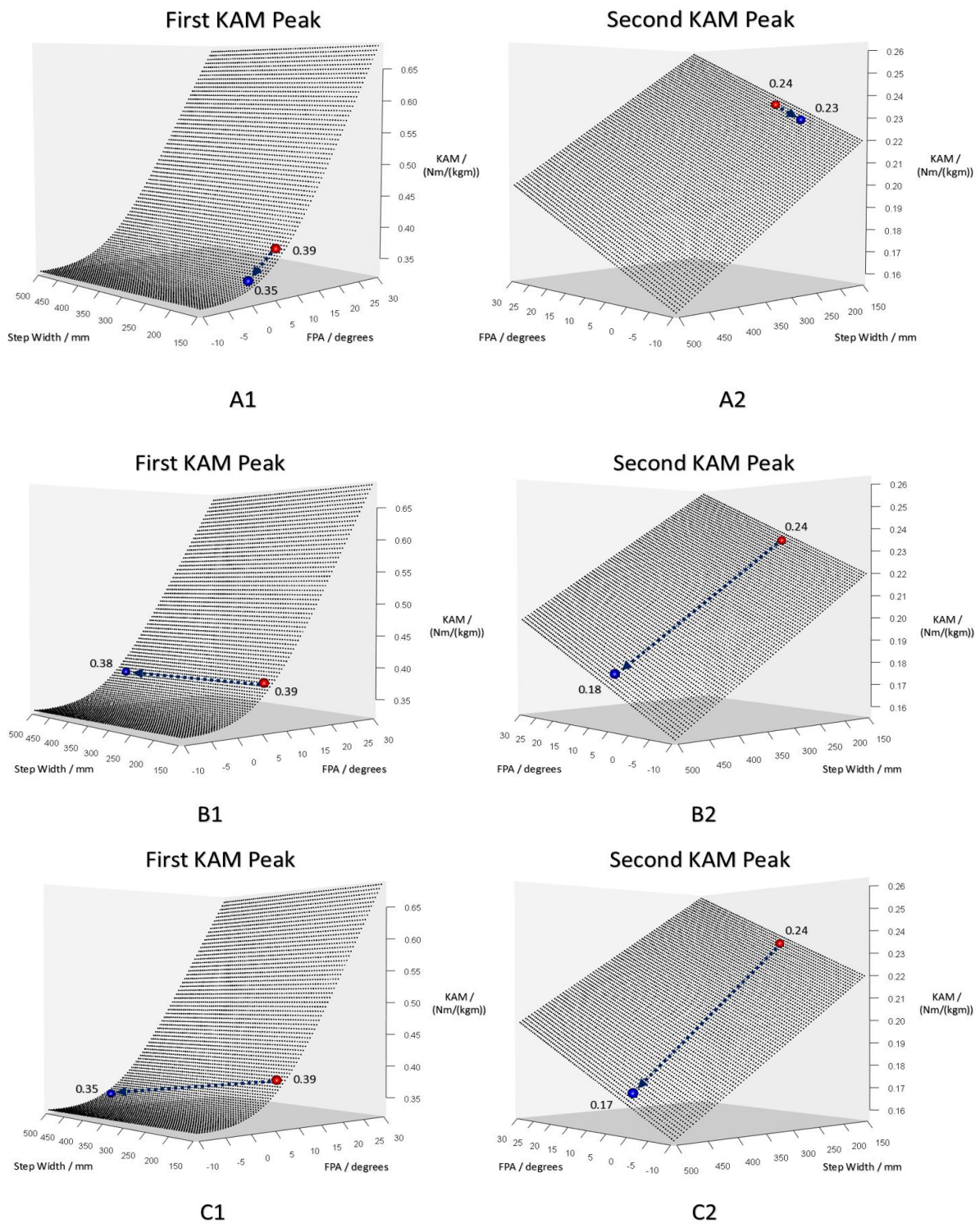

Figure 8.5. Regression models with predicted values (Nm/(kgm) for the first and second KAM peaks for the three new gait modifications. Red points indicate the model predictions for the KAM of normal gait and the blue points indicating the model predictions for the KAM of the new gait. (A1/2) Toe-in with normal SW, (B1/2) Normal FPA with wide SW, and (C1/2) Toe-in with wide SW.

modifications prescribed, we investigated how changes in the FPA and SW of that participant affected the peak KAM compared to normal gait.

Scatter matrices showed that the influence of FPA and SW on peak KAM were participant-specific. This reinforces the need for participant-specific gait retraining strategies explored in the literature [4, 6, 201]. Approximately half the trials showed significant correlations between FPA and KAM peaks, whereas only 10-15% of the trials were significant for SW. One limitation of using Pearson's product-momentum correlation in determining which individuals were promising candidates is that potential nonlinear relationships would not be accounted for. Previous work has shown FPA to have a significant positive correlation with first ($R^2 = 0.419$) and negatively with second KAM peaks ($R^2 = 0.973$) [137]. Correlations for the participant of interest in this study showed significant positive correlations between FPA and first KAM peak ($R^2 = 0.36$, $p < 0.001$) which is consistent with Lynn et al. (2008). However, we also observed a positive relationship between FPA and the second KAM peak ($R^2 = 0.22$, $p < 0.01$), which reiterates the participant-specificity of gait and KAM. Previous research and simulations have suggested that increasing SW reduces peak KAM, implying a negative correlation [128, 130, 194]. However, this only seems to be consistent for the second KAM peak ($R^2 = 0.31$, $p < 0.001$) with no significant correlations detected with the first peak. In order for the KAM to be altered, either the ground reaction force or moment arm needs to be changed. One possibility as to why a correlation was not observed in this participant between SW and the first KAM peak is that other influential parameters (trunk lean, tibia angle etc.) were not controlled for in this study and may have changed when SW was increased, thus masking the underlying correlation. Individuals may change other parameters to compensate for changes in their gait, which was observed in the reduction of walking velocity when individuals internally rotated their foot [127].

Regression models for the first and second KAM peaks for the participant chosen for this study suggested that toeing-in and increasing SW would reduce both KAM peaks. No consensus has been found in the literature with regards to the influence of toeing-in on the KAM. A study in normal teenagers found that toeing-in increased both KAM peaks, with the second being higher than the first [134]. Conversely, it has been reported that toeing-in did not increase early stance KAM but increased it during late stance [137]. Although there is evidence to suggest that internal rotation of the foot reduces the early stance KAM peak in healthy individuals but not in individuals with knee OA [127], the opposite has also been observed

when the FPA is internally rotated by an average of 5° leading to a reduction in the first KAM peak by 13% [136]. From the regression model, it seems that the greatest change in the first KAM peak with FPA is when the angle is greater than approximately 8°. Therefore, our models predict that toeing-in from approximately 7° (normal gait) to 2° would only result in a 10% reduction in the first KAM peak from 0.39 to 0.35 Nm/(kgm) and the second KAM peak barely change from 0.24 to 0.23 Nm/(kgm) when SW is unchanged. The lack of correlation between SW and the first KAM peak as shown by our scatter matrix, is consistent with our first peak regression model being mainly dependent on FPA and not SW. The consensus in the literature about increasing SW during walking has been a reduction in peak KAM [128, 130, 194]. This was consistent with the predictions from the regression models when increasing SW to 500 mm (nominal) with the second KAM peak decreasing from 0.24 to 0.18 Nm/(kgm) and the first peak barely changing from 0.39 to 0.38 Nm/(kgm). Combining modifications of both FPA and SW by walking toe-in at 2° and SW of 500 mm, our models predicted reductions in the first and second KAM peaks from 0.39 to 0.35 Nm/(kgm) and 0.24 to 0.17 Nm/(kgm), respectively. Although our predictions from the regression models were consistent for the most part with trends in literature, they under predicted both the first and second KAM peak values. The prediction error was more apparent for the first KAM peaks especially at wide SW of 500 mm with the absolute differences ranging from 0.19 to 0.24 Nm/(kgm). Even though the exact reason for this disparity is unknown at this stage, we suspect that abrupt changes in the dynamics during the walking phase in Experiment 1 where FPA and SW were varied freely by the participants and the low sample data used in constructing the regression model, especially in the region of wider SW, could have led to these differences. From the model R^2 values of 0.57 and 0.39 for first and second peak, we also expect there to be variability in the prediction against actual measured KAM values.

Toeing-in by approximately 5° alone did not change either the first ($p = 0.994$) or second ($p = 0.179$) KAM peaks. This was also observed in the wide SW cases when the participant walked with normal and toe-in gaits with the first and second KAM peaks being very similar ($p = 0.49$ and $p = 0.84$ respectively). The lack of effect on KAM peaks when walking with toe-in gait has been previously reported [127]. It is possible that a difference may be detected if a greater toe-in angle was prescribed. However, our first peak regression model showed that the reduction in KAM begins to decrease per degree of toe-in after approximately 8°. It seems in this particular participant, of the gait modifications tested, only increasing SW seems to change the peak KAM values. From these results, it is evident that the participant can

significantly reduce the second KAM peak (by 14 - 18%) by walking with a wide SW of 500 mm. However, the trade-off is that it adversely influences early stance peaks, increasing them significantly (by 33 – 37%) and the deviation from normal gait may be too 'awkward' for participants to adopt. Previous studies have shown that the first KAM peak increases by 9.4% to 24% and the second KAM peak decrease by 35.5% to 56% for toe-out gaits [126, 135], we however, observed this in gaits where the SW was increased. It is unclear why the first KAM peak increased when the participant of interest increased their SW contrary to a previous optimization study [128]. Paquette et al. reported that during stair descent with increased SW in healthy older individuals, both the first and second KAM peaks decreased [139]. However, no change in KAM was found when increasing SW in patients with knee OA suggesting increasing SW may not be as influential for symptomatic patients [140]. Paquette et al.'s studies performed during stair descent, therefore it is unknown how changes in SW would translate between healthy and patient populations in the context of level-walking. The ground reaction force and moment arm from the knee joint centre both contribute to the KAM [202]. Therefore, future studies should explore how the ground reaction forces and moment arms change when altering gait parameters.

Adhering to the recommendations of the regression models in this study reduced the second KAM peak but increased the first. The only gait modification which seemed to not increase the first KAM peak was the toe-in gait at normal SW. However, this had no noticeable effect on any of the peaks and therefore poses no known advantages over normal gait. Presently, only the overall peak KAM has been linked to the progression of knee OA [3, 122], which almost always is the first of the two peaks [202, 203]. Very little is known about the effect the late stance KAM peak has on the progression of knee OA. In future studies, we will investigate whether a systematic gait perturbation method, providing equal data points across a range of motion, would change the regression models that were present in this study. At the same time we could investigate controlling for other influential parameters like centre of pressure of the foot [5], tibial angle [4], and trunk sway [4, 138]. A few metrics which will also need to be taken into account in future studies are the knee flexion moment which, when paired with KAM, increase the predictability of the medial contact force [124]; KAM impulse which is also related to the progression of knee OA [204]; metabolic cost of gait modifications because a given gait may reduce medial contact loads but require too much energy expenditure [205, 206]; and pain of a given gait modification. Finally, it remains to be seen whether long-term gait adaptations,

such as those presented here, will have any long-term effects on the health of articular cartilage or the severity of disease progression.

Summary of Conclusions

9.1 Summary of Results

Real-time gait retraining remains a promising non-invasive intervention for reducing medial compartmental loads. The results in this thesis have explored haptic modalities novel to the lower limb which have demonstrated very high direction perception accuracies. We proposed the idea of combining haptic modalities into a single ankle bracelet as a means for miniaturising and making wearable haptics more portable (Chapter 3). Lateral skin stretch is a very effective haptic modality for providing directional cues with certain speed and displacement combinations on specific lower limb loci yielding accuracy rates of up to 100% (Chapter 4). However, lateral skin stretch is not as suitable when the application is for a wearable device and a dynamic activity, as increased mass and tension in the bracelet would be required in order to prevent tactile masking, which would be undesirable for a portable device (Chapter 6). TAM is also an effective haptic modality for the lower limb, yielding high accuracy rates when optimal SD and ISOI combinations were used. TAM is a modality which is easily integrated into a wearable device as it does not suffer from tactile masking due to a reaction force, thus it did not only yield high accuracy rates and low response times during static settings but also during walking (Chapter 5, 6).

A novel multi-modal haptic ankle device was proposed which combined binary vibrotactile feedback with TAM, using a four ERM actuator array. The haptic device was evaluated in a gait retraining task for FPA and SW. Individuals were able to intuitively retrain both parameters using the prototype in a short training session (< 15 mins) (Chapter 7). FPA and SW were chosen as they were gait parameters which are influential in reducing the KAM but also natural enough that gait modifications would be more easily adopted than schemes involving trunk sway. In the final chapter (Chapter 8), we proposed a data-driven method of reducing peak KAM by only altering FPA and SW. The KAM peaks from the data-driven model of the participant that was trialled, was different from the calculated KAM measured. The gait recommendation given was able to reduce the late stance KAM peak but adversely affected the first peak. Despite the undesirable result in this study, this disparity between model and experimental KAM peak values is suspected to be because of the experimental methodology when the participant was given freedom to vary his FPA and SW; leading to involuntary secondary gait kinematics which differ to when the participant was walking at a consistent and non-varying gait.

9.2 Conclusions

In conclusion, the research presented here demonstrates that it is possible to retrain two gait parameters using a single wearable haptic feedback device. This improves on previous systems in terms of portability where two separate haptic devices would have been needed on different body segments. A data-driven approach of obtaining gait modifications, namely FPA and SW, which reduced the peak KAMs was presented, despite only being able to reduce the late stance peak and not the first. Future work discusses experiments and necessary steps required to further the field of real-time gait retaining.

9.3 Future Work

Future research should continue to explore multi-modal haptic feedback in wearable devices especially in the lower limb. Continuing to question how future devices can be further simplified and miniaturised without sacrificing effectiveness as a feedback device also warrants further study. Stemming from Chapters 4 and 5, future research should investigate the numerous gait parameters which have the potential to be modified, coming up with better gaits through the use of models, and creating models which consider other important aspects of gait, for example, accounting for any adverse secondary changes which occur. Lastly, the results presented in this thesis could be implemented to the broader field of motion guidance, rehabilitation, and high-performance sports.

Despite lateral skin stretch being deemed unsuitable for our particular wearable haptic application, that does not prevent applications in static, grounded settings from applying the results we presented in Chapter 4. Environments where a grounded reaction force can exist, for example: in vehicles, aircrafts, and stationary robotic orthoses; lateral skin stretch can be a new way of providing proprioception or directional feedback cues to the lower limb. Development of new ways of providing skin stretch; whether through the use of new actuators or novel mechanical design which eliminates tactile masking, could make lateral skin stretch devices more portable and a viable form of wearable haptics in the future. Improvements to the current ankle haptic device could continue, for example, exploring the use of linear resonant actuators instead of current ERM motors for improved perception and power consumption in a portable device. Reducing the number of motors in the bracelet from four, and comparing whether or not the reduction changes user perception of TAM is also of interest. Presently, the ankle bracelet only elicits two different directional cues. Research for additional cues could allow a

single device to retrain more than two gait parameters. It seems wearable haptics, as it stands, is mature enough for use outside of laboratory and clinical settings, although much remains in the area of kinematic sensing before real-time gait retraining can be applicable. The main challenge that needs to be met is developing wearable devices with sensors and sensing algorithms which give comparable millimetre position accuracy like the motion capture systems which exist in laboratories.

The results in this thesis show that gait modifications involving only FPA and SW were not able to reduce the first KAM peak for that particular participant. As mentioned earlier, this is likely due to the experimental methodology of varying the gait parameters. Gait kinematics during FPA and SW variations are expected to be different compared to normal walking gait, and there are possible secondary effects due to changes in other gait parameters, such as trunk sway. Future research should explore whether or not a standardised or more systematic gait varying methodology during data collection would create more accurate regression models. Increasing the number of data points would also reduce uncertainty in the models and give a clearer picture as to whether this data-driven approach can be a reliable method of optimising gait in the future. For real-time gait retraining to have higher chances of being adopted outside of laboratory settings, the gait modifications must not deviate too much from 'normal looking' gait. Therefore, modifications involving only FPA and SW were used here. With more accurate regression models and experimental data, it might be possible to determine whether or not FPA and SW modifications alone can give comparable results to invasive interventions, such as high tibial osteotomy surgery. Current studies have focused on reducing the first KAM peak, but more recently, the applied knee flexion moment has been recognised as an important inverse dynamic load that can lead to increased joint contact force at the knee. Future gait recommendations should account for secondary effects of gait modifications like metabolic cost, loading in other joints and muscles, and their long-term effects. This will give researchers a greater understanding of what the best recommendations are in light of different existing trade-offs.

The focus of this thesis has been towards real-time gait retraining for slowing the progression of medial compartmental knee OA. The research presented here, however, could be extended into other avenues such as stroke rehabilitation, high performance sports training for athletes, and injury prevention. With hardware and algorithms constantly improving, it is

possible in the near future that real-time wearable motion guidance will no longer be just another science experiment but a very real solution which can be integrated into our daily lives.

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