Optimization of portable electronically-controlled needle-free jet injection systems

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Abstract—Jet injection is a process by which a fluid drug is delivered through the skin in the form of a high-velocity jet. Powering jet injection using a controllable actuator, such as a moving-coil permanent magnet motor, offers many advantages, but to date has required large and heavy injection systems to provide the required power and control bandwidth. In order to minimize the size of the injection system, we developed a scaling model for jet injection systems powered by permanent magnet motors, giving the optimal actuator mass as a function of jet velocity, injection volume, motor efficiency, and energy storage density. We combined this model with an existing electromagnetic model to confirm the predicted scaling relationships and find optimal actuator designs. On this basis, we designed an injection system for 50 μL volumes, including a compact power amplifier and control system, and verified its performance by performing injections into pig skin. The total mass of the injection system was 578 g, with a 178 g handpiece. This model illustrates fundamental relationships that govern the design of any jet-production device powered by linear electric motors, for jet injection or other applications.

Index Terms—Drug delivery, biomedical electronics, design optimization, electromechanical systems, actuators.

I. INTRODUCTION

JET INJECTION is a promising technique for the needle-free administration of injectable drugs in the form of a high-velocity liquid jet. The jet is produced by rapidly pressurizing a volume of a liquid drug formulation, typically to pressures between 20 MPa and 100 MPa, so that it can be expelled through a narrow orifice, typically between 100 μm and 500 μm in diameter [1]. This technique holds promise for reducing pain, avoiding needle phobia, and increasing the productivity of health care staff, but to date these benefits have rarely been realized [2].

Traditionally, jet injectors have been purely mechanical devices, using the triggered release of a spring or compressed gas to drive a piston that pressurizes the fluid. The evolution of the pressure (and thus the jet velocity) over time in such devices is fixed by the design of the energy release mechanism: spring-powered devices begin at a high pressure that decays linearly as the drug is delivered, while compressed gas devices can provide a relatively constant pressure. These jet speed profiles are able to deliver drugs through the skin, but devices generating them are also often associated with “wet” injections (i.e. drug remaining on the skin surface), bruising at the injection site, and/or pain [2]–[4].

Recently, jet injection systems based on electronically controlled actuators have been developed [2]. These systems are able to precisely tailor the jet speed profile for each injection, potentially enabling a reduction in pain and other adverse events, as well as allowing the same injector to be used for a wide variety of different injections. To date, controllable jet injectors have used either moving-coil permanent magnet electromagnetic actuators [6]–[8] or piezoelectric actuators [9], [10]. Of the two, only electromagnetic actuators, as shown in Fig. 1, are capable of delivering single doses comparable to those delivered via traditional needle administration.

However, the electromagnetic injectors developed to date have been larger and heavier than spring-powered injectors, and have relied on a tethered connection to a massive power and control system. This arises because the injection process demands control of the motor with a bandwidth of over 100 Hz while delivering 5 kW or more electrical power to the actuator, for up to 100 ms. For instance, the injector in [7] uses an actuator with a mass of 1.5 kg to deliver 100 μL, while the system reported in [8] uses a 450 g actuator to deliver 300 μL. The power amplifier employed in both systems (AE Techron LVCS5050) has a mass of 35 kg; a self-contained amplifier for the system in [7] was constructed [11], but had a comparable mass and occupied a volume of 250 L. By comparison, a typical spring-powered device, the Comfort-In injector (Mika Medical), has a mass of only 167 g, and delivers 500 μL. Smaller, but uncontrolled, capacitor discharge power systems have also been attempted. The device described in [12]...
was designed as a self-contained hand-held unit including a capacitor for power, but did not incorporate a control system or sufficient energy storage for full-volume injections; similarly, Chen et al. [6] describe an electromagnetic injector powered by capacitor discharge, but do not control the injection or provide details regarding their actuator or power system design.

With the actuator and power supply required to operate at very high power density, design optimization of the system can help to reduce its required size and weight. To date, this has not been thoroughly explored for systems employing moving-coil actuators acting against fluid loadings. Design optimization has been employed extensively for valve actuators [13]–[15], but without consideration for the energy source or power electronics. Actuator designs intended for jet injection have been developed [8], [16], but these designs have not been informed by motor models or optimized.

One potential path to reducing the size and mass of electromagnetic injectors and their control systems is through reduction of the injection volume. To this end, in this paper we present a model linking the injection volume of an electromagnetic jet injector to the mass of the actuator and energy storage system it requires. Guided in part by this model, we describe the development of a lightweight, portable, controllable jet injection system, optimized for a delivery volume of 50 µL. First, we describe a scaling model and optimization procedure that minimizes the combined mass of the energy storage and actuation components of an injector system. We then present the design of a compact power supply and control system capable of up to 7 kW output power and 250 J energy delivery, sufficient to drive an optimized injector as well as an existing jet injection system [5] for the 50 µL volume. Performance of the controller is verified using this injector mechanism to deliver injections into porcine tissue. We then present the design of an injector mechanism optimized for this delivery volume, and demonstrate its injection performance.

II. DESIGN THEORY

The greatest challenge in controllable jet injection using electromagnetic actuators is in the supply of the very high power required by the actuator during an injection. While moving-coil permanent magnet actuators have a fast response and are simple to drive, they offer relatively low efficiency. A particular challenge is that the power required is directly related to the actuator mass [5]: reduced power requires increased actuator mass \( M \). However, reduced power consumption also reduces the size and mass of the components that store the energy for the injection. Here, we discuss the basic physics of electromagnetically-actuated jet injection, and present a scaling model that predicts the optimal actuator size to minimize the combined mass of the motor and energy storage system.

A. Injection physics

Ignoring losses due to fluid viscosity, the jet formation process can be described using Bernoulli’s equation. While viscous losses can be significant in some circumstances [17], for fluids similar to water the losses are small and do not alter the basic proportionality between pressure and velocity. Based on this inviscid assumption, the power dissipated in the actuator \( P_d \) is given by

\[
P_d = \frac{\rho^2 V_{\text{max}}^2 v^4}{4 K_m^2 L^2},
\]

(1)

where \( K_m \) is the motor constant (in \( N/\sqrt{W} \)), \( L \) is the maximum stroke of the piston (and thus the motor) driving the fluid drug, \( \rho \) is the density of the drug solution, \( v \) is the jet velocity, and \( V_{\text{max}} \) is the maximum drug volume.

The power output \( P_o \) depends only on the fluid, the jet velocity, and the orifice diameter \( d \):

\[
P_o = \frac{\pi \rho v^3 d^2}{8}.
\]

(2)

Typically, the power output in a jet injection system is sufficiently small compared to the power dissipated in the motor that it can be ignored.

In order to design the power system for an injector, it is also necessary to know the total energy required by the injection. Working from the injection volume \( V \) and jet speed, the duration \( T \) of a single-phase injection can be found as

\[
T = \frac{4V}{\pi vd^2}.
\]

(3)

The energy dissipated in the actuator \( E_d \) can then be calculated as

\[
E_d = P_dT = \frac{V_{\text{max}}^2 \rho^2 v^4}{\pi d^2 K_m^2 L^2}.
\]

(4)

After the initial penetration of skin, it is possible to deliver the remaining fluid at a lower jet velocity [2], [8], [10]. The total energy required for a two-phase injection like this can simply be found as the sum of the energy required for each phase. This can be managed most expeditiously by choosing a modified injection time \( T \) such that it gives the equivalent energy for a single-phase injection at the highest jet velocity employed.

B. Examples

The injection system described in [5] comprises an actuator with a motor constant of 3.2 \( N/\sqrt{W} \) and a stroke of 30 mm, delivering a fluid with the density of water at 150 m/s through a 300 µL ampoule’s 220 µm orifice. This basic model implies a power dissipation of 1.2 kW, for an output power of just 64 W. If it delivers its full 300 µL volume at 150 m/s, an energy of 65 J is required.

In practice, the power and energy required are increased due to friction and viscous loss in the ampoule. In [5], it was found experimentally that an input power of approximately 7 kW was required to achieve a 150 m/s jet velocity. Typically, we find that jet velocities suitable for tissue penetration (≈ 120 m/s) can be obtained with input power comparable to that required for an ideal jet velocity of 200 m/s (i.e. 20 MPa). This leads to an injection power (for 300 µL) of 3.9 kW, and an energy of 260 J.
Due to the fundamental scaling laws observed by electromagnetic systems [18], [19], if a permanent magnet linear motor is scaled without altering the proportion of its components, the motor constant varies in proportion to the square root of the actuator mass: \( K_m = \bar{K}_m M^{1/2} \), where \( K_m \) is the motor constant proportionality factor. With this scaling condition, the stroke \( L \) varies like any other linear dimension: \( L = \bar{L} M^{1/3} \), where \( \bar{L} \) is the factor of proportionality. (Scaling in such a manner also requires the design of the fluid ampoule to be adapted to the motor mass.) We can thus say that \( P_d = \bar{P} M^{-5/3} \), where \( \bar{P} = \frac{\rho^2 V_{\text{max}}^2 v^4}{4K_m^2L^2} \). (Scaling relationships hold true, it is informative to use the electromagnetic model to generate power-optimized designs over a wide range of motor sizes. This can be done by performing constrained minimization on the electromagnetic model, with injection power (1) as the objective function and the motor mass constrained to a fixed value. The motor performance varies as a function of the coil position; to ensure good performance over the entire stroke, the injection power is calculated at both mid-stroke and end-stroke, with the worse of the two values used as the objective. This also leads to a conservative estimate for the required injection energy. Fixed radial gaps of 0.1 mm and 0.4 mm are dictated outside and inside the coil, respectively, to allow for free coil motion and to provide space for mechanical support of the coil.

The results of the optimization, performed using the SQP algorithm [20], are shown in Fig. 3 for actuator masses from 10 g to 10 kg; a 300 \( \mu \)L injection volume and 200 m/s injection velocity have been assumed. There are two distinct optimal designs: a local optimum, characterized by a short radial magnet and a strong variation of performance over the motor stroke; and the global optimum, which has a very long radial magnet and equal performance at the end of stroke to the middle of the stroke. The local optimum needs approximately 15% more power for small motors, closely approaching the global optimum for larger motors. There is some variation of the proportionality factor with motor mass, amounting to approximately a twofold variation over three orders of magnitude of mass.

The long radial magnet of the globally-optimal design may pose problems in motor assembly, due to the strong mutual repulsion between the magnet segments. The construction of designs at the local optimum is significantly more straightforward, and so only designs of this type were pursued for further study.

Equipped with the proportionality factor, it is now only necessary to determine the power and energy density associated with the energy storage to design an optimized injection system. The power and energy requirements for injection suggest two possible classes of energy storage component: photoflash-grade electrolytic capacitors, which are energy-limited in this
application, and electrochemical supercapacitors, which are power-limited.

While each capacitor type requires comparable mass and space to meet the requirements of jet injection, photoflash capacitors offer a better impedance match to practical actuators, with moderate voltages and currents compared to the low voltages and high currents associated with supercapacitors. (There are also safety and reliability benefits to operation away from the power limits of the components.) For example, a typical photoflash capacitor (Suntan CD17, 360 V, 1500 µF) has an energy density of 950 J/kg when fully charged. Photoflash capacitors have sufficiently high power densities (> 1 MW/kg) as to always be energy-limited for jet injection.

Using this energy density, we can find the optimum motor mass for both the injection volume in [5] (300 µL) and the 50 µL volume proposed in this work. In all cases, we will assume a 200 m/s jet velocity for purposes of power calculation, but a 120 m/s jet velocity to calculate the injection duration $T$. The results for a range of injection volumes are shown in Fig. 4; the best fit to this scaling is $M_{opt} \propto V^{1.077}$, slightly below the expected power law due to the variation in $P$ with mass. The power required to perform injections using optimized systems varies relatively little with volume, ranging from 2 kW to 3.5 kW for volumes from 10 µL to 10 mL.

For a 300 µL injection system, the optimum actuator mass is 301 g, approximately 35% higher than the nominal actuator mass of 222 g implemented in [5]. This leads to a total system mass of 480 g: 180 g of capacitors are used, rather than the 270 g required to operate the existing injector. Thus, inclusion of the power system in the optimization shows that the efficiency lost in trying to miniaturize the size and mass of the motor alone cancels out the mass savings. However, the system masses are comparable; the optimum is a shallow one, and there is a wide range of feasible solutions close to the optimal design with similar performance.

For a 50 µL injection system, the model suggests a nominal actuator mass of 45 g and 24 g of capacitors, for a total system mass of 69 g. In each case the power required for injection is similar: 2.6 kW for the 300 µL system, and 2.2 kW for the 50 µL system.

III. INJECTOR SYSTEM DESIGN

A. Miniaturized injector

Due to magnet availability and manufacturing constraints, a slightly larger actuator than the optimal design was developed, with a nominal mass of 62 g and key dimensions given in Table I. The impact of this deviation from the optimum is reduced by the much higher efficiency of this larger actuator, however, with a change in the estimated system mass of only 13%. The radial magnet section was achieved using six uniformly-magnetized wedge segments, closely approximating an ideal radial magnetization. According to the electromagnetic model, this motor should require a maximum power of 1.5 kW to perform an injection, and have a maximum motor constant (at the mid-point of travel) of 3.2 N/√W. The motor was wound with 245 turns of 27 AWG wire, for a nominal coil resistance of 2.2 Ω. Accounting for the additional mass of the coil bobbin and the structural components of the motor, its total mass was 89 g.
This actuator is coupled to a stainless steel ampoule, with a bore of 2.4 mm and fitted with a precision 200 µm orifice (O’Keefe Controls ZMNS-8-M3.5-SS-BN). A stainless steel piston with o-ring seals is used to drive the injection fluid through the orifice. The piston position is sensed using a linear potentiometer (ALPS RDC1014A09). The complete mechanism, including the ampoule and the position sensor, has a mass of 142 g. To illustrate the potential use of the device, it was enclosed in an ergonomic case, as shown in Fig. 5, with a total hand-piece mass of 178 g.

B. Power supply and controller

With the actuator design parameters established, we set out to design a power electronic and motor control system suited to powering both the existing 300 µL injector and an optimized 50 µL injector. The basic parameters for the larger injector were given in Section II-B; in order to achieve a nominal injection pressure of 20 MPa, our existing actuator requires power to be delivered in the form of 30 A of current at 130 V. Furthermore, in order to accurately control the jet speed profile a controller with at least 100 Hz full-power bandwidth is required [8]. An optimized 50 µL injector requires a similar amount of power, but for a much shorter time and thus less energy.

Photoflash capacitors (Suntan CD17, 3 × 1500 µF) were chosen as the energy storage component, with a total mass of 306 g and maximum stored energy of 290 J at 360 V. The capacitors are charged from a low-voltage DC power input via a flyback converter, as shown in Fig. 6, based around the Linear TL3750 controller and Coilcraft DA2033 transformer and capable of charging the capacitor array in approximately 20 s from a 15 V supply. To provide additional safety margin, the capacitors are only charged to a maximum voltage of 320 V, storing 230 J. A diode bypass of the capacitor charging circuit is also provided, so that the motor can operate directly from the power supply when the capacitors are not charged.

Power is controlled and delivered to the actuator using an H-bridge, comprising IGBTs (Infineon IRGS30B60K) with anti-parallel SiC diodes (Cree C3D10060G) and switched at a frequency of 20 kHz in a locked-antiphase pattern. Gate drive is provided using high-voltage level shifters (Infineon IR21141), while current is measured synchronously twice per PWM period using a level-shifted current sense amplifier (Infineon IR2177). Additional low-impedance DC bus capacitance (Vishay MKP1848, 15 µF, 700 V) is provided to protect the bridge during regenerative operation. The system is rated for a maximum measurable current of 50 A, and thus a maximum peak power delivery of 16 kW.

System control is accomplished using an ARM Cortex-M4 microcontroller (ST Micro STM32F405RG) operating at a clock frequency of 168 MHz. The microcontroller measures the voltage on the photoflash capacitors, the motor current, and the motor position (via a potentiometer transducer). Standalone operation is supported, though for testing purposes injection data are transmitted between the controller and a PC via a TTL serial connection.

The control electronics as a whole can be implemented on 3 PCBs to fit into a 90 mm × 45 mm × 25 mm volume, with a mass of 94 g, as shown in Fig. 7. This corresponds to a peak rated power density of 158 W/cm³. For bench testing purposes, a controller on a single PCB was also constructed.

The controller firmware supports open-loop duty-cycle control, open-loop voltage control, closed-loop position control, and closed-loop current control of the injector mechanism, all performed twice per PWM cycle for a 40 kHz control loop rate. Current control is accomplished using a PI compensator; position control employs a lead compensator, with the zero at 100 Hz and the pole at 400 Hz, in conjunction with integral action and feed-forward compensation. Once tuned, this controller provides a bandwidth of approximately 200 Hz. Fixed-point saturating arithmetic is used to implement both controllers. Feed-forward compensation is based on the
differentiated as a function of time. The commanded piston velocity, with calibration performed by ejecting water underwater using voltage control.

A simplified block diagram (omitting saturation and anti-windup) for the position controller is shown in Fig. 8. In this block diagram, the commanded position is \( u_c \), and the actual motor position is \( x \); \( a, b, \) and \( c \) are the calibrated feedforward terms, \( K_L \) and \( K_I \) are the lead and integral gains, respectively, \( \omega_z \) and \( \omega_p \) are the zero and pole frequencies for the compensator, and \( V_{cap} \) is the capacitor voltage. The compensators output a control effort in terms of voltage, which is normalized by the capacitor voltage to determine the required PWM duty cycle \( D \) of the amplifier.

![Block diagram](image)

**Fig. 8.** Block diagram for the position controller, with the feedforward path at the top, the lead compensator in the center, and the integral compensator at the bottom. The control effort is normalized by the capacitor voltage to find the PWM duty cycle; the plant includes the H-bridge as well as the actuator and drug ampoule.

**IV. PERFORMANCE VERIFICATION**

**A. Power supply and controller**

Performance of the power and control system was first demonstrated using a larger injector mechanism developed in [5]: an actuator with a stroke length of 30 mm, a coil resistance of 4.4 \( \Omega \), and a motor constant of 3.2 \( N/\sqrt{V} \) coupled to a commercially-available 300 \( \mu \)L jet injection ampoule (Injex part #100100). The stock piston associated with the ampoule was replaced by a rigid steel piston with o-ring seals to reduce the mechanical compliance of the system.

In Fig. 9, the power system is shown driving the injector under closed-loop position control, to deliver a 100 m/s jet.

![Graph](image)

**Fig. 9.** Actual (solid line) and commanded (dashed line) piston position are shown for a position ramp (nominal jet velocity of 100 m/s and piston acceleration of 76 \( m/s^2 \)) with a water-filled ampoule.

The feed-forward element in the controller is essential [8] in reaching the desired jet velocity rapidly; a brief (5 ms) acceleration phase at the start of the trajectory reduces velocity overshoot. The lead compensator causes the slight overshoot at the end of the injection waveform; this overshoot is repeatable and can be compensated for in trajectory design.

The power system was also used with this injector to perform 50 \( \mu \)L injections into tissue, under open-loop duty cycle control. Skin from the chest and upper torso of pigs ranging from 9 to 12 weeks of age was harvested in accordance with the University of Auckland Code of Ethical Conduct for the Use of Animals for Teaching and Research. Tissue was frozen at -80 \( ^\circ \)C immediately post-harvest, thawed overnight at 4 \( ^\circ \)C prior to use, and allowed to equilibrate for 30 minutes immediately prior to injection. A blue dye (1.8 \% Brilliant Blue FCF; Queen New Zealand Pty. Ltd.) was used as the injection fluid, to aid in visualization. After injection, the tissue was immediately re-frozen at -80 \( ^\circ \)C, then sawed in half for imaging.

Fig. 10 shows the results of one such injection, performed using a constant PWM duty cycle of 75 \%, including the piston position and nominal volumetric jet velocity. The injector reaches a steady-state jet velocity of approximately 130 m/s; this choice of control input also gives a brief high-velocity impulse at the beginning of the injection to assist penetration, created by the compliance of the fluid and ampoule. The injection shown required 35 J of energy from the capacitors, for an average power consumption of 3.5 kW.

![Graph](image)

**Fig. 10.** 50 \( \mu \)L injection performed using the portable amplifier with the large injector and 300 \( \mu \)L ampoule from [5]. It has been controlled to remain near the surface, using a 75 \% duty cycle and 10 ms duration. The piston position (solid line) and nominal jet velocity (dashed line) for the injection are shown.

**B. Miniaturized injector**

The properties of the actuator for the optimized 50 \( \mu \)L injector are compared to the expected values in Table II. The motor constant over the range of motor positions is shown in Fig. 11 and compared to the predictions of the electromagnetic model; the shape is very similar to that predicted, with a

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**Table II. Motor Constant for Different Motor Positions**

<table>
<thead>
<tr>
<th>Motor Position (mm)</th>
<th>Motor Constant (N/\sqrt{V})</th>
</tr>
</thead>
<tbody>
<tr>
<td>10</td>
<td>3.2</td>
</tr>
<tr>
<td>20</td>
<td>3.5</td>
</tr>
<tr>
<td>30</td>
<td>3.8</td>
</tr>
</tbody>
</table>

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consistent, slight performance reduction largely caused by iron saturation.

The combined optimized injector mechanism and power system were used to deliver injections into tissue under open-loop duty cycle control, as shown in Fig. 12. This injector mechanism had a greater mechanical compliance than the larger system, resulting in a more pronounced initial high speed phase. The velocity achieved in this phase was very sensitive to the applied power, as shown by the different depths achieved in these two injections (at 25% and 27% duty cycles). The 25% duty cycle injection required a total energy of 19 J, for an average power consumption of 1.9 kW.

V. DISCUSSION

The scaling model shows, perhaps counter-intuitively, that a jet injector operating with an energy-limited power system is best designed for approximately constant power dissipation over a wide range of injection volumes. This has important implications for the motor drive electronics, which therefore have limited ability to scale with the rest of the injection system. Smaller energy storage capacitors directly reduce the size of the charging circuit required, but this only uses 20% of the mass of the electronics in the system as built. Closer coupling of the electronics to the storage capacitors might allow for a reduction of the DC bus capacitance in the motor driver, which occupies nearly half of the volume and one third of the mass of the electronics. The use of wide-bandgap (gallium nitride or silicon carbide) MOSFETs in place of the IGBTs in the motor driver would enable a further size and mass reduction; however, even with these improvements, a fully-controllable motor driver would likely be larger and heavier than the actuator for injection volumes below 50 µL.

The actuator optimization process does lead to a single acceptable optimum, but it is important to note that the optimum is relatively shallow, particularly with regards to the use of larger actuators with smaller energy storage systems. This may provide some leeway for reducing the size of the motor drive electronics, though the high voltage employed tends to limit their minimum size.

At the other end of the size range, the model conclusively rules out the development of hand-held injection systems with injection volumes larger than approximately 0.5 mL, for which the actuator alone has a mass of about 500 g. The use of high-performance supercapacitors in place of photoflash electrolytic capacitors for the energy store could lower the actuator mass somewhat, but only with the attendant costs and reliability risks associated with operating them at their power limits. Instead, other approaches will be needed to achieve high-volume injections. For instance, a mechanical transmission can be employed to reduce the force (and power) required for the initial phase of a two-phase injection [21]. Alternatively, motor designs that permit independent adjustment of the stroke length and motor constant can be employed [22]. Both these approaches have drawbacks, however, with reduced controllability and increased mechanical complexity.

The injector system itself performs similarly to the model predictions, and has demonstrated both high-performance control and successful depth-controlled injection. Simple control strategies were used to perform the injections, so as to aid in understanding the basic injection processes; work is ongoing to determine the effects of different sequences of injection velocity upon injection depth and retention of the injected fluid. Even a simple, constant-duty-cycle control is sufficient to control injection depth, and uses the stored energy more efficiently than a simple capacitor discharge circuit would. The
assembled injector hand-piece, tethered to the power system, is comfortable and easy to use, and will be used in future clinical trials of small-volume jet injection.

VI. CONCLUSION
The jet injector scaling and optimization model presented in this paper establishes concrete limits on the practical range of injection volumes achievable with controllable jet injectors. These insights can be used to guide future development of devices of this type; for large injection volumes, it is clear that alternative mechanisms will be needed. We have demonstrated that the use of small injection volumes allows for the construction of small, lightweight, hand-held injectors with dynamic control over the injection process, and that the model accurately describes the performance of such an injector. In this process, we also developed a compact pulsed-power amplifier that is able to drive the injector under closed-loop position control with feedforward compensation. These outcomes and methods can be used to guide the design of any self-contained, portable, high-power electromagnetic actuator system.

REFERENCES

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