Rehabilitative Knee Orthosis Driven by Electro-Rheological Fluid Based Actuators

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Abstract - This work aims to demonstrate the feasibility of using Electro-Rheological Fluid (ERF) actuators in orthotics, creating a new breed of rehabilitation devices. ERFs are fluids that experience dramatic changes in rheological properties, such as viscosity or yield stress, in the presence of an electric field. Using the electrically controlled rheological properties of ERFs, compact actuators with an ability to supply high resistive torques in a controllable and tunable fashion, have been developed. This study involves the design, fabrication and testing of an ERF based knee orthotic device and the innovative ERF actuators it uses. The knee orthotic is achieved through a standard brace design with a polycentric hinge and gear system. Coupled to this are two Flat-Plate ERF actuators, given that name for their characteristic set of parallel flat plates allowing for actuation of the fluid. A full model describing the field dependant torque output of these actuators is presented along with a full detailed description of the device design. The overall knee orthotic system is designed to resist up to 25.4% of an average human knee's torque abilities and be controlled in real-time. The goal of this work is to provide a much more efficient means of rehabilitation over the average orthotic, while matching the proficiency of rehabilitation machines, all in a smaller, simpler, and more cost efficient design.

Index Terms – Rehabilitation Robotics, Smart Orthoses, Electro-Rheological Fluids, Compact Actuators.

I. INTRODUCTION

This research aims to prove that using ERF actuators to make active orthotic devices, can bridge the gap between passive orthotics and rehabilitation machines. Both categories have their obvious advantages as well as drawbacks. Orthotics being smaller in size, inexpensive, and readily available, are regrettably very inefficient if at all useful in terms of rehabilitation. Rehabilitation machines, superior in their ability to provide optimal levels of resistance and control in rehabilitation regimes, are excessively large, costly, and their widespread use is hindered by these weaknesses. This work attempts to combine the positive aspects of both worlds while downplaying their faults, in a new breed of rehabilitation devices.

An orthotic by strict definition is a specialized mechanical device that supports or supplements weakened or abnormal joints or limbs. The majority of these devices can be categorized as passive, meaning the resistance or support they provide is not changed in real time. The Sports Medicine Committee of the American Academy of Orthopedic Surgeons has further classified these types of braces, specifically used for the knee, into four categories: prophylactic, rehabilitative, functional and patellofemoral. All provide stability, apply precise pressure, and/or help maintain alignment of the knee joint at set constants.

Some of the more innovative designs allow torsion to be applied at the knee joint and new technology has further improved their efficiency by allowing the torque to be adjusted. However, the lack of real-time abilities is a significant downside for these devices that limits their overall effectiveness in rehabilitation. The inclusion of active components has been a widely accepted method of improving upon this deficiency.

This seemingly small addition has considerable drawbacks though. The application of traditional active elements increases the overall size, cost, weight, and other related characteristics. Equally important are the concerns with control and sensory feedback, which would also be considered necessary with the addition of active components. All these combined, along with the obvious goals of making the systems as efficient and beneficial to an individual during rehabilitation as possible, force their designs to go beyond that of a portable orthosis, and more so a machine.

In terms of rehabilitation, the most effective methods known today are these rehabilitation machines. They are commonly used for rehabilitating and strengthening patients, subjects, and athletes while providing quantitative measurements of their performance. They provide high resistive and sometimes assistive forces, while providing a unique tailoring of the rehabilitation regime to nearly any individual. This ability dramatically increases their proficiency as a rehabilitation tool. Their services have been limited to primarily only physical therapy offices though, as a direct result of their shear size, weight, and cost.

In this paper we investigate the feasibility of using Electro-Rheological Fluid (ERF) actuators in orthotics, creating a new breed of rehabilitation devices. While matching the proficiency of rehabilitation machines, will be smaller, simpler, and more cost efficient. Using the electrically controlled rheological properties of ERFs, compact actuators with an ability to supply high resistive torques in a controllable and tunable fashion, have been developed. This study involves the design, fabrication and testing of an ERF based knee orthotic device and the innovative ERF actuators it uses. A full model describing the field dependant torque output of these actuators is presented along with a full detailed description of the device design.
II. ELECTRO-RHEOLOGICAL FLUIDS

Electro-rheological fluids (ERFs) are fluids that experience dramatic changes in rheological properties, such as viscosity, in the presence of an electric field. Willis M. Winslow first explained the effect in the 1940's using oil dispersions of fine powders [1]. The fluids are made from suspensions of an insulating base fluid and particles on the order of one tenth to one hundred microns (in size). The volume fraction of the particles is between 20% and 60%. The electro-rheological effect, sometimes called the Winslow effect, is thought to arise from the difference in the dielectric constants of the fluid and particles. In the presence of an electric field, the particles, due to an induced dipole moment, rearrange into a more organized manner, or form chains along the field lines as it is shown in Fig. 1. These chains alter the ERF’s viscosity, yield stress, and other properties, allowing the ERF to change consistency from that of a liquid to something that is viscoelastic, such as a gel. ERF’s generally respond to changes in electric fields in a matter of only a millisecond or two. Good reviews of the ERF phenomenon can be found in [2-4].

![Particle suspension forms chains when an electric field is applied.](image)

Under zero field conditions an ERF is generally characterized by a simple Newtonian viscosity. When subjected to high electric fields, the ERF alters its state from Newtonian oil to a non-Newtonian Bingham plastic. As a Bingham plastic, the ERF exhibits a linear relationship between stress and strain rate, just like a Newtonian fluid, but only after its particular yield stress has been exceeded. Before that point, it behaves as a solid. This shear stress behavior of an Electrorheological fluid is described most simply by the well known Bingham Model:

\[
\tau = \tau_y + \mu \dot{\gamma}
\]

Where \(\tau\) is the shear stress, \(\tau_y\) is the yield stress, \(\mu\) is the dynamic viscosity and \(\dot{\gamma}\) is the shear strain. The dot over the shear strain indicates its time derivative, the shear rate.

The yield stress, \(\tau_y\), and the dynamic viscosity, \(\mu\), are the two most important parameters that affect the design of ERF-based devices. The dynamic viscosity, \(\mu\), is generally determined by the base fluid with some field dependency which is often neglected when using the Bingham Model. The field-induced yield stress, \(\tau_y\), is generally a function of the electric field strength and considered shear rate independent. The Bingham plastic model has been widely used to predict the post-yield behavior of ERF, i.e. the behavior of the ERF when flowing. A review of the material compositions for ERF patents can be found in [4]. As an example fluid, the properties for an ERF designated LID 3354S, manufactured by Smart Technology Ltd. [5], is as follows. LID 3354S is an electro-rheological fluid made up of 35%, by volume, polymer particles in silicone/fluorolube base oil. It is designed for use as a general-purpose ER fluid with an optimal balance of critical properties. Its physical properties are: density: \(1.46 \times 10^3\) (kg/m³), viscosity: \(110\) (mPa·sec) at 30°C; boiling point: > 200°C; flash point: > 150°C; insoluble in water; freezing point: < -20°C. The field dependencies for this particular ERF are:

\[
\tau_y = C_y E^2
\]

\[
\mu = \mu_0 - C_{\mu} E^2
\]

where: \(\mu_0\) is the zero field viscosity and \(C_y\) and \(C_{\mu}\) are constants that can be determined experimentally.

Control over a fluid’s rheological properties offers the promise of many possibilities in engineering, especially actuation and control of mechanical motion. Devices that rely on hydraulics can benefit from ERF’s quick response time and reduction in device complexity. Their solid-like property in the presence of a field can be used to transmit forces over a large range and have found a number of applications. A list of many engineering and practical applications of ERFs can be found in [6].

III. ERF ACTUATOR

A. ERF Actuator Concept

To both actuate and make optimal use of the ER fluids characteristics, a flat plate actuator concept was developed (see Figure 2 - left). The concept uses a rotating electrode plate placed between two stationary electrodes. The three components are separated only by a thin layer of fluid and applying an electric field across the gap alters the fluid’s properties. More specifically, the fluid’s yield stress is increased. When the middle plate is in motion, the higher yield stress corresponds to an increased shear stress on the electrode’s surfaces.

The actuation of the viscous fluid is consequently creating a resistive torque on the rotating shaft. By manipulating the strength of the electric field sent through the fluid, the torque can be easily controlled, turning this simple concept into an actuator.

Maximizing the surface area that moves through the viscous fluid is the best way to increase the torque/force output from an ERF-based actuator, so multiple parallel rotating electrode plates are used. This allows for maximum shearing surface area while maintaining a compact overall volume for the actuator.

Two different sets of conductive plates are fabricated and oriented in alternating fashion, with one set rigidly attached to the actuator to prevent any movement, while the other set is attached to a rotating shaft. These alternating plates serve as the positive and negative electrodes that generate the electric field to actuate the ERF that fills the gap between the plates. The whole assembly is neatly placed within a housing, filled with ERF, and sealed to complete the assembly (Fig. 2, left two pictures).

B. Flat Plate Actuator Model

The performance of the flat plate (FP) actuator concept is directly related to three key factors. These are the geometry of the actuator, the input voltage sent to the electrodes, and the properties of the ERF itself.

The geometry components of the actuation model are all parameters of the flat-plate electrodes. These are the inner radius of the electrode plates (\(r_i\)), outer radius of the electrode plates (\(r_o\)), number of plates, and the gap width...
between plates (d). Fig. 2, right two pictures, below shows these four variables graphically.

The torque output equation of the flat-plate actuator using these variables and the specific fluid properties are:

$$T = 4\pi N \left( \frac{r_i^3 - r_f^3}{3} \right) \tau_y + \mu \frac{r_i^4 - r_f^4}{4d} \omega$$

(4)

Where: N is the number of moving plates, $\tau_y$ is the yield stress of the fluid, $\mu$ is the viscosity of the fluid, and $\omega$ is the angular velocity of the plates. Every type of ERF is composed of a different composition of suspended particles in a fluid base and thus has its own unique behavior and properties. Therefore, each ERF has its own characteristic relationship between the two and it must be known in order to derive a complete and accurate model for the actuators. The fluid chosen for this project was ERF-3356S, an ERF created by Smart Technologies Ltd. After testing of the fluid and determination of its properties, the final modelling equation for the FP actuator using this fluid is:

$$T = 4\pi N \left( \frac{r_i^3 - r_f^3}{3} \right) (0.179E^2 + 0.253E + \tau_f) + \mu \frac{r_i^4 - r_f^4}{4d} \omega$$

(5)

Where: $\tau_f$ is the no-field frictional yield stress term characteristic to each specific actuator, $\mu$ is the dynamic viscosity of the fluid and is equal to 187 [cp], and E is the electric field governed by the relationship:

$$E = \frac{\text{Voltage [kV]}}{\text{d [mm]}}$$

(6)

Derivations of the basic FP model and details on testing to determine the fluid properties can be found in [6, 7].

IV. DEVICE DESIGN

The proposed design consists of three major subsystems – the actuator, support/frame, and the gearbox and other miscellaneous components. Each subsystem includes several components of varying complexity. In general, all of the components were designed with strength and safety in mind and optimized for regular testing and high-stress torture testing. All CAD models and mechanism analysis were performed using Pro-E Wildfire. The complete CAD model for the ERF Knee Orthosis is shown in Fig. 3.

A. Design Specifications

For the final ERF Orthotic device to be feasible and superior to presently available devices, a number of specific goals had to be met. The final design had to be complimentary to the complex biomechanics of the knee joint, as well as comfortable and non-restrictive for the user to wear. It had to withstand the high torque capabilities of the knee joint, while providing controllable resistance during extension and flexion. Employment of ERF actuators were to be used to provide this unique, variable, real-time resistance of the device that is largely lacking in presently available orthotics. A design goal of 25% of a healthy knee’s torque abilities was set for the performance requirements of the actuator. Finally, future considerations for marketability were also to be considered, meaning the overall cost for the device would need to be competitive with presently used rehabilitation devices.

The highest torque a knee is capable of occurs during knee extension, at an angle of approximately 90 degrees. The maximal force an average knee is capable of producing at the ankle at this angle is 344.63 Newtons [8]. The ankle is approximately 0.35 meters below the knee on an average sized adult. Thus, the approximate maximum torque an average knee is capable of is 120.62 Newton meters. The average speed of the human leg during walking is 1.77 meters per second, meaning an angular velocity, $\omega$, of approximately 5.07 radians per second at the knee joint would be a good estimated velocity for the actuator. The inner radius of the plates was set to a constant 20.0 millimeters in order to allow for a strong axle structure within the actuator. From knowledge gained with previous ERF actuators (see [6], [7], [9], and [10] for these), suggestions from Smart Technology Ltd., and a design goal to minimize the overall size of the system, it was opted to set the gap size to the smallest deemed possible, $d = 1.0$ millimeters, and set a reasonable maximum voltage at 4.25 kilovolts.

With all those in place, the design of the actuator was simplified down to the selection of two variables, N and $r_o$. These were directly related to the overall dimensions of the actuator, deciding its height and width, respectively. Now, knowing the relative dimensions of an adult human leg, and a goal of making the orthotic practical, the overall dimensions of the actuator had to be unobtrusive to the motion of the leg and limited to a moderate size. Unfortunately, the use of just one actuator resulted in awkwardly huge dimensions for N and $r_o$. Similarly, the employment of two actuators, one on either side of the knee, still resulted in excessively large actuator dimensions. It was found that if the torque from the leg could be geared down as well as using two actuators, the goal of 25% of the human legs capabilities would be possible. A gear ratio of 1.67 was selected for reasons of availability and easy attachment to the brace, making the required output torque for each actuator now only 9.03 Newton meters. It is important to note that although the inclusion of a gear system meant the torque requirements for each actuator were decreased; it also meant the angular velocity experienced by each would then be increased. When adjusting the angular velocity variable accordingly in Equation (5) it is seen that a simple doubling of the angular velocity only increases the output torque by less than one percent, meaning its effect is negligible.
With \( N \) set at 8 rotating plates and \( r_o \) at 45 millimeters, the actuator was theoretically capable of outputting 9.17 Newton meters of torque. This was a practical dimension for the actuators and meant the whole system was capable of resisting 25.4% of the average human legs capabilities. Below is Table I summarizing the requirements of the system, the limitations of the design, and the stated actuator parameters that match these requirements.

<table>
<thead>
<tr>
<th>Design Goal:</th>
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<tbody>
<tr>
<td>% Human Capabilities</td>
<td>25%</td>
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<tr>
<td>Orthotic Torque Desired</td>
<td>30.16 [N m]</td>
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<table>
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<th>Set Actuator Limitations:</th>
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<td>Gap size ( (d) )</td>
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</tr>
<tr>
<td>Inner Radius ( (r_i) )</td>
<td>20.0 [mm]</td>
</tr>
<tr>
<td>Angular Velocity ( (\omega) )</td>
<td>5.07 [rad/sec]</td>
</tr>
<tr>
<td>Voltage ( (V) )</td>
<td>4.25 [kV]</td>
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<tr>
<th>Designed Actuator Parameters:</th>
<th></th>
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<tr>
<td>Number of Plates ( (N) )</td>
<td>8</td>
</tr>
<tr>
<td>Outer Radius ( (r_o) )</td>
<td>45 [mm]</td>
</tr>
<tr>
<td>Max. Actuator Torque</td>
<td>9.17 [N m]</td>
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<table>
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<tr>
<th>System Characteristics:</th>
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<tbody>
<tr>
<td>Gear Ratio</td>
<td>1:1.67</td>
</tr>
<tr>
<td>% Human Capabilities</td>
<td>25.4%</td>
</tr>
</tbody>
</table>

### B. Orthotic Design

As opposed to designing and fabricating a brace specific for this project, it was opted to rather use an off-the-shelf knee brace proven to meet all the requirements necessary for this project. The brace chosen was the Don Joy 4TITUDE™, seen below in (Fig. 4).

![Fig. 4 Don Joy 4TITUDE™ brace (left); CAD model of Don Joy 4TITUDE™ brace (with gearbox) (right).](image)

The frame provided an excellent platform to base this design on, constructed from a lightweight aluminum it was exceptionally rigid and well designed, so it was expected to perform favorably with the addition of a resistive component. It was available in many sizes and incorporates a proven polycentric hinge design. The polycentric hinge is a mechanism which mimics the natural rotation and gliding motion of the knee joint. Two axes are connected by a linkage and mated with elliptical gears. The motion this produces can be seen in the path trace in Fig. 5, which is almost perfectly matched to the motion characteristic of the knee joint.

![Fig. 5 Kinematics of polycentric hinge with path line in red.](image)

Mechanically, the joint on the Don Joy 4TITUDE™ brace had \( \frac{1}{4} \) in. shafts of rotation, with axes 1” apart. The rotation axis was offset below the frame center. The shafts are connected via thin aluminum cover plates on both sides of the frame. These plates could easily be removed and allowed easy adaptation of the necessary gearbox/actuator subsystem.

A four-point attachment system was used in the Don Joy 4TITUDE™, allowing for rigid coupling to the leg and minimizing brace migration. This attachment system, along with the polycentric hinge, is a patented innovation by the company, termed “Four Points of Leverage”. It is detailed on their website as follows; “The Four Points of Leverage combine to successfully apply a net differential posterior force, or constant force to the tibia, which reduces [instabilities]” [11]. The attachment system was more then sufficient for securing the orthotic to a human knee and combines comfort with easy application and removal of the device.

### C. Actuator Design

The actuator design specific to this knee orthotic was actually a fourth generation ERF based actuator design. For more details on previous actuator development, design and testing, see [6], [7], [9], and [10]. It possessed very large torque capabilities and full 360 degree motion, both areas of weakness with previous ERF actuator designs. It is important to note that during the design process, it was expected that for actual fabrication of the actuator, a Rapid Prototyping method would be used. This allowed the design to be impressively detailed and precise.

The assembly of the new FP actuator consisted of a lid and the case body. The lid incorporated a secure box-style method of attachment (Fig. 6), by way of six M3 socket head machine screws and acted as the actuator’s mount to the system. It had cut out volumes for the brushes and the wire path, seen in Fig. 6, left picture. A Teflon spring seal was used for the output shaft hole to ensure durability, longevity, and low friction. The overall dimensions of the lid were: a diameter of 118.5 millimeters and 20.64 millimeters high.

The actuators casing, shown in Fig. 6, measured 38.1 millimeters high, with an outer diameter of 127 millimeters, and inner diameter of 107.95 millimeters. On the exterior of the casing, mounts for an optical encoder were placed to supply accurate readings of rotational position. The encoder requires access to the transmission shaft, so the shaft actually extends from both sides of the actuator. Teflon spring seals were also used in the casing’s shaft hole to prevent fluid leakage.

![Fig. 6 Actuator lid, detail of wire patch and brush cutouts (left) and gearbox mounts with recess for snap ring (center) and Actuator casing (right).](image)

A three-point attachment system was used for connecting a \( \frac{1}{4} \) steel shaft to the centrally mounted rotating electrodes (Fig. 7, left and middle pictures). A set of support surfaces would be available for each plate to fit in between, and would couple the electrode plates rigidly to the rotating shaft via three mounting screws. One row of support surfaces (from the three star design) was removed to allow for the integration of custom machined copper spacers (Fig. 7,
right picture) which formed a direct rigid electrical conduction pathway between electrodes.

Fig. 7 CAD model of rotating central shaft (mounting for rotating plates) (left & middle); CAD model of copper spacer 1.81mm x 6mm x 5mm (right)

The steel main output shaft would run through and be completely enclosed by the rotating shaft component. As long as an electrically non-conductive material was used for this component, such as the epoxy resin used in rapid prototyping processes, a requirement of complete electrical isolation would be met. The central rotating plates (Fig. 8, left) were therefore designed to easily slide in to and attach to this rotating component, to be made out of a highly conductive material such as copper. They were locked into place with 3mm round head screws, as mentioned.

Fig. 8 CAD models of rotating electrode plate (left); and stationary electrode plate (right)

A large diameter brush/commuter system (Fig. 9, left) was designed to allow for full 360 degree motion while maintaining a solid electrical connection through which the plates could be energized. The commuter would attach to the rotating shaft component and be made of a highly conductive material also, such as copper. Trinity Racing brushes, usually used in remote control cars, were chosen to be modified to mate for use by altering the contact radius to ½” diameter to mate with the ½” diameter commuter. These would be placed into the lid with miniature compression springs that apply the necessary force for low electrical resistance contact between the brushes and the commuter.

The mountings for the fixed plates were incorporated into the actuator housing. The case mounts (Fig. 9, right) are shaped in such a way as to allow them one degree of freedom when inserted into the case shell; they slide freely to help component alignment but stay in the proper angular position. The interior components of the actuator would have the unique ability to be completely assembled as well as serviced outside of the case, simply sliding in and out of the case.

Fig. 9 Actuator brush and commuter components (copper and blue color), electrodes (green), and rotating shaft (purple).

Gap size was also a very critical factor of the FP design; if the plates skewed too close, arcing would occur, if they drifted too far apart, performance would decrease. To keep the plates in position, maintaining a constant 1 millimeter gap, Teflon guides were added to the design. A 0.8 millimeters groove, to match the plate thickness, was to be machined in 1.6 millimeter Teflon hollow medical tubing. These grooved guides would be epoxied to the inner and outer edge of the fixed and rotating plates. These Teflon guides, riding along each other on the edges, would help maintain the gap width between the 0.8 millimeter thick plates (chosen for their satisfactory rigidity at this thickness) while producing very little friction. In addition they would provide insulation at the edges of the plates, eliminating the arcing that might occur at the sharp edges of the plates.

The full CAD models of the actuator assembly can be seen in Fig. 10. In Fig. 11, the interior components of the FP actuator are shown with more contrasting and realistic colors. Any confusion in the design will most likely be clarified in these. The views display all major components, with the color coding as follows: electrically active components are copper colored, spacers are blue, and the rapid prototyped epoxy components are earth shades.

Fig. 10 CAD model of interior components of actuator assembly (left); The completed actuator assembly, less the lid and commuter, with a transparent casing (center); CAD model of exterior components (right)

Fig. 11 Fabricated case with o-ring seal, brush-commuter system, spacers, and electrode plates (left); electrode plates with rapid prototyped mounts (right)

D. Gearbox Design

The gearbox’s (Fig. 12) primary function was to transmit and multiply the torque output of the actuators. Steel spur gears with a gear ratio of 1 to 1.67 were chosen based on the three parameters. These were basic gear availability, the calculated actuator torque requirements, and on the need to have the gears on the same 1” centers as the hinge pivots of the Don Joy 4TITUDE™ knee brace. A 0.75” and 1.25” pitch diameter gears were chosen. A 3/32” keyways and machined aluminum adaptors (which also acted as spacers) kept the gears securely attached to the shafts and hinge.

Fig. 12 CAD model of assembled gearbox

The gearbox case halves would be attached to the support plate on the interior side of the hinge by 3 millimeter socket head machine screws placed in positions that would not interfere with the hinge operation. ¼”-28 low profile machine screws inserted through the hinge axis directly into
the transmission shafts, would further stabilize the hinge mechanism.

The actuator would need to be coupled to the gearbox in a way that provided reliable torque transmission and convenient installation and removal of the actuator. This requirement would be satisfied by soldering a half shaft into the previously mentioned bore; the actuator output shaft would be milled to mate with this “half moon” shape (see Fig. 12). When inserted, the shaft of the gearbox and actuator would lock together in rotation.

E. Additional Considerations

To make this design an effective and feasible one, there are a number of other components, which needed to be considered and included as part of the orthotic. These included such factors as the sensors, control board, customizing of the system to the user, and a power source. Sensors integrated with closed loop control are what would allow the orthotic to have an advantage over presently available orthotics. Sensors would have to provide a method of determining the angle of the knee joint and the torque being produced by the knee. As mentioned, an encoder was included into the design; however, no torque sensors were included. Given that this was simply a first prototype design for proof of concept, this was deemed acceptable, but is a requirement for any future work with this concept.

Additionally, a number of other components were not specifically included in the final design for the actual orthotic, but they all could be replaced temporarily by basic lab equipment and computer hardware. A control board would need to be mounted either on the orthotic or on a belt worn by the user and would include a micro-processor, D/A and A/D boards, motion controllers, sensor interface boards and a modem card for radio communication with a remote control station. The ability to communicate with external hardware or PC’s and a closed loop control providing robustness and redundancy, is what would allow for the construction of optimal rehabilitation regimes.

Given the fact that a very low level of electric current (on the order of milliamps) is needed to activate the ERF elements, the orthotic is expected to operate using a relative small and lightweight battery. These batteries could be easily placed on a belt or somewhere on the orthotic as well and quickly recharged. Due to the high voltages that the system requires though, special considerations in safety using such items as circuit breakers, fuses, or trips, would also have to be included in the electronics to ensure that the device did not cause any bodily harm.

V. Prototype

To complete the project a prototype of the designed device was built (see Fig. 13-15). It was fully functional and was built exactly as described in the previous design sections. The Don Joy 4TITUDE™ knee brace was donated by the company Don Joy Orthopedics. It was slightly disassembled and machined in house to allow for attachment of the gearbox and actuator. The actuator casing, internal components, and lid were rapid prototyped using the 3D Systems Viper 2000i2 machine. The electrode plates and spacers were contract out to be cut out of copper plates. Due to the large number of high precision plates needed this was chosen to ensure a consistent accuracy in fabrication of all of them. The device used approximately 187.5 milliliters of fluid. Below are several images of the prototype and close-ups of some of the individual parts (Fig. 13-15).

Fig. 13 Fabricated case with o-ring seal (top left); CNC machined electrodes with rapid prototyped mounts (top right); fabricated rotating shaft with steel output shaft and commuter installed (bottom left); actuator shaft with rotating plates attached (bottom right)

Fig. 14 Fabricated gearbox (left); Inner hinge (center left); Attached Actuator casing made with slots so inside plates are visible (center right); Fabricated actuator attached, filled with fluid, and encoder mounted (right)

Fig. 15 First version prototype of the ERF driven knee Rehabilitation Orthosis (left actuator casing is made with slots so inside plates are visible, right actuator is filled with fluid)

Fig. 16 Theoretical vs. Experimental Torque of final designed/fabricated actuator

So far tests were performed to verify the capabilities of the actuators. Since two identical actuators were used, the verification of one of these actuators would be theoretically as accurate as creating a duplicate of a human knee joint for the purpose of testing the whole device. The average torque output of the actuator at each voltage was plotted and compared to Equation (5), the predicted theoretical equation’s results. Fig. 16 was the result and the two plots show a very close resemblance. The accurate results therefore suggest, the proposed system is capable of the forces desired and is predictable through Equation (5). More importantly, this verified the success of the project in that the system has been proven to meet its design goals.
VI. CONCLUSIONS AND FUTURE WORK

This work was started with the argument that it is possible to build a superior, low cost, and compact ERF driven orthotic that can bridge the gap between passive orthotics and rehabilitation machines. The design process of the system involved the meshing of standard, mechanical solutions and novel, new age ones. Throughout the work it has become evident that all these design goals have been addressed and satisfied.

Performance requirements were all met. An anticipated goal of 25% an average human knee’s abilities was set as a reasonable torque for proof of concept. With torque capabilities of 30.64 [N m], the system just slightly exceeds the design requirements. In addition, the final test results on the fabricated actuator verify the accuracy of the models and the predicted performance of the system.

The complexity of the human knee joint posed interesting issues with the overall design of the device. Instead of a single degree of freedom system, the proper biomechanics required the use of a multi-degree of freedom joint. This was addressed through the use of proven orthotic components, more specifically a Don Joy knee brace and its polycentric hinge.

The ability of rehabilitation machines to function in real-time, is what separates them from their less efficient counterparts, orthotics. The inclusion of ERF actuators met this design criterion. They are easily controlled, due to their field dependant torque outputs, and can react on the order of milliseconds. Therefore, the truest form of real-time control, closed loop control, is a realistic possibility with the designed ERF Orthotic.

The obvious smaller size of the device eliminates any concerns with storage, portability, and weight that rehabilitation machines are faced with. Furthermore, a relatively low cost is an attractive advantage of the device. The constructed device came just short of $1000 to build. With all the electrical and sensory components that need to be added to the device for a final functional and marketable product, it is estimated to cost approximately $3500. A state of the art, computer controlled Isokinetic Machine, such as the Biodex System 3 Pro, can be bought for $44,900.00. Size and cost issues are therefore clearly overcome with ERF based orthotics.

Although sufficient to demonstrate the feasibility of an ERF orthotic, the device built does require many alterations in order to be a fully functional rehabilitation device. Future work aims to improve upon most of the shortcomings of the device. These include improvement to its resistive torque capabilities, sensory-based control, safety, and selection of materials.

Besides improving the design and control of the system, such aspects as clinical trials would need to be performed. Injuries would obviously be the first most obvious application, but knee replacement surgery is another closely related situation where rehabilitation of the knee is necessary and ideally suited for this device. Helping stroke patients recover their mobility is another interesting area with promising applications for this device.

A number of other applications exist where this same concept can be used as well. One is nearly identical in application and extends the technology to elbow rehabilitation. The elbow is a nearly identical cryonival joint and an ERF orthotic poses all the same advantages over current state devices as the knee application does. Another interesting application could be in microgravity compensation. Extended trips in space are marked with significant loss in the bone and muscle mass of astronauts. To maintain their body’s strength, they are required to exercise regularly and even then it is insufficient. By applying a similar device as this orthotic to the joints of a suit of an astronaut, the effects of gravity can be simulated on the body, thus preventing all the effects of weightlessness.

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