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# Clinical potential and design of programmable mechanical impedances for orthotic applications

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**SUMMARY**

A person with a moderate or severe motor disability will often use specialised or adapted tools to assist their interaction with a general environment. Such tools can assist with the movement of a person's arms so as to facilitate manipulation, can provide postural supports, or interface to computers, wheelchairs or similar assistive technologies. Designing such devices with programmable stiffness and damping may offer a better means for the person to have effective control of their surroundings.

This paper addresses the possibility of designing some assistive technologies using impedance elements that can adapt to the user and the circumstances. Two impedance elements are proposed. The first, based on magnetic particle brakes, allows control of the damping coefficient in a passive element. The second, based on detuning the P-D controller in a servo-motor mechanism, allows control of both stiffness and damping. Such a mechanical impedance can be modulated to the conditions imposed by the task in hand. The limits of linear theory are explored and possible uses of programmable impedance elements are proposed.

KEYWORDS: Mechanical impedance; Clinical potential; Motor disability; Orthotic application

## 1. INTRODUCTION

Although the term 'impedance' is widely used in electrical engineering it has equal significance in mechanical engineering where it represents the dynamic characteristics of a mass-spring-damper system. Lawrence[1] and Hannaford[2] have used the results of mechanical impedance theory in conjunction with 2-port theory to describe the dynamic response of master-slave telemanipulators. Linear models for the operator, mechanism and environment allow some indication of how the remote environment will be perceived through the limitations of the master-slave mechanism.

The concept of mechanical impedance has also been used to gain a better understanding of the mechanics of the muscular skeletal system and Hogan[3] postulated that human movement can be considered as a set of programmed impedance characteristics commanded by higher centres that determine the approximate endpoint of movement but contain no explicit information about the path of movement. Hogan suggested that human movement is achieved by the central nervous system establishing appropriate muscle stiffness that remains constant for the bulk of an uninterrupted movement and which alone determines the movement trajectory. This theory has received further development from Mussa-Ivaldi[4] who mapped out the impedance fields in frogs and postulated that four Gaussian fields could be used to specify a full range of limb movements.

The concept of mechanical impedance has also been applied to haptic displays[5]. The term haptic display has come to mean electro-mechanical linkages that attempt to emulate the surface shape and features of an imaginary object based on parameters that are stored in a computer. Most haptic environments attempt to enforce the concept of passivity, that is the requirement that over an undetermined period of time the system should dissipate mechanical power. Thus a haptic display would allow an operator to open a virtual door, but not allow the door to vibrate with an increasing amplitude perpendicular to the hinge. The design criteria for haptic displays are considerably more arduous than designing programmable impedance elements for clinical use since most haptic interfaces use as a bench mark standard that of replicating the stiffness of objects commonly encountered in a physical environment whereas programmable impedance for clinical use need only simulate a stiffness comparable with that associated with human joints.

The concept of mechanical impedance has much to offer rehabilitation robotics and powered orthotic mechanisms where there is a trend to design flexible and potentially lower cost linkage systems. Although such systems will be unable to meet the needs of all individuals who experience difficulty with manipulation, these linkages may be a suitable and cost effective solution for individuals who have well defined needs for assistance with manipulation. Individuals who need to enhance existing manipulative skills, such as assisting residual movement, damping tremors or stabilising their arms in suitable locations in the environment, would use a powered orthotic mechanism to support their arm along with an appropriate interface to command the mechanism impedance and end point. For individuals who need to transfer and amplify movements from a non traditional body site some type of telemanipulator is required. Under both circumstances the assumption is that the individual has adequate proprioception to be able to utilise the principle of extended

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physiological proprioception[6], or that there is a clinical justification for promoting arm movement. When this is not the case the telerobotic device would be required to respond to a series of user commands and operate essentially unsupervised while a task is in progress. In such a case a high level computer controller would set the stiffness and damping of the mechanism. Where a mechanisms supports, enhances or transfers human movements the programming commands could be preset, determined by the user or selected by a computer according to the needs of the task. Irrespective of the mode of operation it will be assumed that the mechanism mass is minimised and that power can be transmitted without loss from the environment and the human through to an appropriate actuator.

## 2. CLINICAL JUSTIFICATION

Use of programmable impedance elements could be considered to be appropriate in three broad areas, posture support mechanisms, rehabilitation mechanisms and mechanisms to assist or replace function. Each of these areas would be associated with different populations of users. Posture support mechanisms are appropriate for any individual having weak muscles or poor muscle tone and would be assigned to the person for as long as was needed. Rehabilitation mechanisms would be expected to be used on a short term basis such as in a hospital setting and would be used by the individual during their recovery, and mechanisms assisting or replacing function would include a wide range of rehabilitation technologies such as rehabilitation robotics, joystick interfaces to wheelchairs, or orthotic mechanisms.

Individuals who might benefit would include people recovering from cerebral vascular accidents such as stroke, or head injury, individuals with high level spinal cord injury, individuals with muscle weakening caused by muscular dystrophy, or motor neuron disease, individuals with intention tremors caused by conditions such as Parkinson's disease, multiple sclerosis, Friedriech's ataxia, or any of the cerebral palsies, and individuals with limited range of movements such as arthrogryposis, or rheumatoid arthritis. Further details of the aetiology, and the USA figures for incidence and prevalence for of some these conditions are given in Stanger and Cawley[7] and in Reinkensmeyer et al.[8]

### *2.1 Clinical approach*

Several existing devices have been proposed that use fixed mechanical damping to assist the functioning or recovery of people with motor impairments. Hendriks et al.[9] experimented with using fixed damping in the joystick interface to a wheelchair to suppress the tremor of the operator. A commercial device to assist eating is marketed by Michaelis Engineering (Buxton, England). This linkage uses a settable damper mechanism on a two degree of freedom manipulator to allow a person with movement tremor to move a spoon between the plate and his or her mouth. Both these examples have mechanical damping characteristics that are fixed at either the time of manufacture or that are mechanically adjusted just before the device is used.

A passive orthosis mechanism was built and tested by Rosen et al.[10] This mechanism had two planar and one rotational degrees of freedom and used magnetic particle brakes with position encoders to achieve a controlled endpoint resistance. Rosen terms this device a controlled-energy dissipation orthosis (CEDO) and reports results for a tracking task along with positive anecdotal comments from the subject group who were individuals with unspecified intention tremors.

Work by Rahman et al.[11] studied the use of a novel mechanism for gravity compensation of a powered orthotic device to allow an individual to have a greater range of motion. This work considered an approach where the mechanism is initially passive and unpowered but designed so that actuation can be added at a later stage. The concept of a variable stiffness impedance makes this possibility relatively straightforward.

Reinkensmeyer et al.[8], Lumm et al.[12] and others considered powered orthotic mechanisms for rehabilitation of individuals following a stroke. This type of mechanism was conceived to move the individual's arm through a range of movements that can be preprogrammed by the therapist, can utilise any residual capacity of the affected arm or mirror movements of the unaffected arm. From a safety consideration it is desirable that the stiffness of such a device be adjusted to the person's movement characteristics.

There are several implementations of rehabilitation robotic mechanisms that provide a variety of functions and levels of control for a person requiring assistance with manipulation[13], but few of these provide any level of programmable stiffness. Work by Harwin et al.[14] demonstrated a head operated telemanipulator mechanism where a limited control of impedance was possible, but like most work in telemanipulators the design goal was to achieve maximum apparent stiffness between the master and slave robots without compromising stability. Provision of programmable impedance in simple linkages designed for telemanipulation tasks will facilitate using robots both in a telemanipulation mode to transfer a person's abilities to a remote manipulator and provide greater tolerance to discrepancies in the environment where the robot is expected to work autonomously according to the user's commands.

## 3. DESIGN OF PROGRAMMABLE MECHANICAL IMPEDANCE ELEMENTS

### *3.1 Implementation of a passive damping element with magnetic particle brakes*

**3.1.1. Theory.** The use of fixed mechanical damping components is very common in mechanical engineering design, with applications ranging from controlling vibrations to over-damping mechanisms to reduce risk of damage. Although many commercial damping elements have been designed to allow a mechanical adjustment of the damping coefficient, few allow the damping element to adapt to the surrounding circumstances. There are notable exception such as the area of active suspension, and in this field the concept of an electrically adjusted damping

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coefficient was developed in cohorts with research on Electro-Rheological Fluids (ERF)[15]. A similar programmable change in viscosity is possible in fluids that exhibit Magneto-Rheological characteristics. Component finding some novel applications are magnetic particle brakes that use the pseudo viscous properties of magnetic particles in a way that is analogous to applications of Magneto-Rheological Fluids. A typical particle brake will have a circular plate attached to a shaft and housed in a chamber containing a dry stainless steel powder. Windings on either side of the plates allow a magnetic field to be established and the powder particles form along chains between the plate and the housing[16]. The resistive torque that can be opposed by a particle brake is proportional to the density of chains and hence to the current that is flowing in the windings that create the magnetic field.

The fact that a particle brake can only provide a resistive torque makes it attractive for clinical applications since there are far fewer safety considerations. A strictly passive damping element was built in the configuration shown in Figure 1. An analogue differentiator was used in preference to a tachometer since the underlying concept was that a position measurement would be necessary for other applications such as movement assessment, wheelchair joystick control or a higher level control system. This position signal is differentiated and passed to an analogue multiplier. Thus the gain of the element can be controlled from some other source such as a computer or the processed information from other sensors in the system.

To simplify the power driver circuit the signal is then full wave rectified before being delivered to the particle brakes. Because the inductance seen at the terminals of the magnetic particle brake is a function of the current flowing in the windings, a current controlled power amplification stage was used.

**3.1.2. Results.** A test bed was built based on the 24 volt B15 brake manufactured by Lake Placid Industries, (Lake Placid, New York). The brake has a rate maximum holding torque of 0.17 Nm which is achieved when the coil winding current is 0.22 amps. The relationship between the brake torque ($T$) considered in the positive sense, and current ($i$) can be approximated by the formula

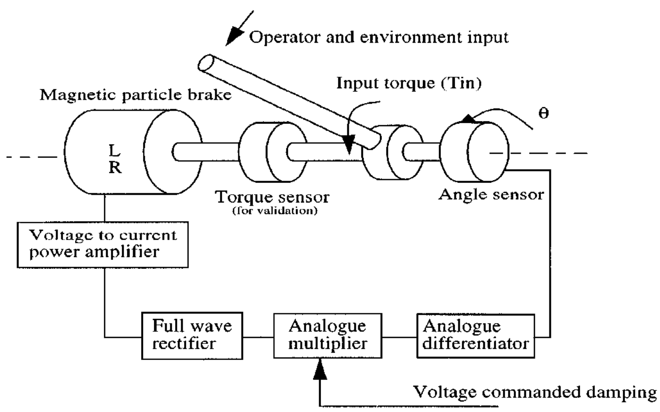


Fig. 1. Variable Damping element based on particle brakes

$$T = \begin{cases} 0 & i < 0.034 \\ 8.89i - 0.3 & i \geq 0.034 \end{cases}$$

There was no attempt to compensate for the discontinuity at the origin in the electronics. The coil inductance is greater than 0.4 Henrys and is probably a function of current. The coil resistance is 105 Ohms.

A torque sensor, the ATI mini from Assurance Technologies Inc. (North Carolina, USA) was used to measure the input torque and the resulting angular velocity was estimated from the output of the differentiator. The degree of damping was controlled by a digital to analogue converter and data was logged onto a computer. The damping element was tested under three conditions: no gain, a gain corresponding to mid-range of damping and a gain that caused the brake to come close to saturation. Figure 2 shows the range of damping coefficients that could be achieved with this arrangement. These results show the nonlinear characteristic of the brake at low currents (less than 0.04 amps), a consequence of the assumption that torque and current are linearly dependent, and illustrates a high degree of hysteresis. Hysteresis under the no gain condition can be assumed to be due to the friction in the mechanism, the hysteresis at higher gains is inherent in the characteristics of the brake.

### *3.2 Implementation of an electro-mechanical impedance with a detuned P-D controller*

**3.2.1 Theory.** Although there are several ways of producing variable stiffness mechanisms, the approach most suitable to clinical mechanisms is to use an electric motor in a closed loop position control configuration such that an external torque results in a spring/damper like behaviour. Although pneumatic mechanisms have some attraction in clinical mechanisms, the upper boundary on stiffness is limited by the compressibility of air thus potentially limiting the useful range of achievable mechanism stiffness. To achieve a programmable mechanical stiffness based on position servo mechanisms requires the motor, linkage and position sensor to operate in a linear region over a wide range of conditions.

Figure 3 shows the arrangement of a single degree of freedom controllable impedance element. Referring all inertia and viscous elements through any gearing to the motor shaft and summing torques gives

$$T_{in} + T_m = J\ddot{\theta} + B\dot{\theta}$$

where $T_{in}$ represents the forces applied by both the operator and the environment, $T_m$ is the motor torque, B and J are the lumped damping and inertia respectively and include any components from a linearised model of the operator, and $\theta$ is the shaft angle.

The motor is modelled as

$$V_{in} = K_T\dot{\theta} + Ri + L\frac{di}{dt}$$

and

$$T_m = iK_T$$

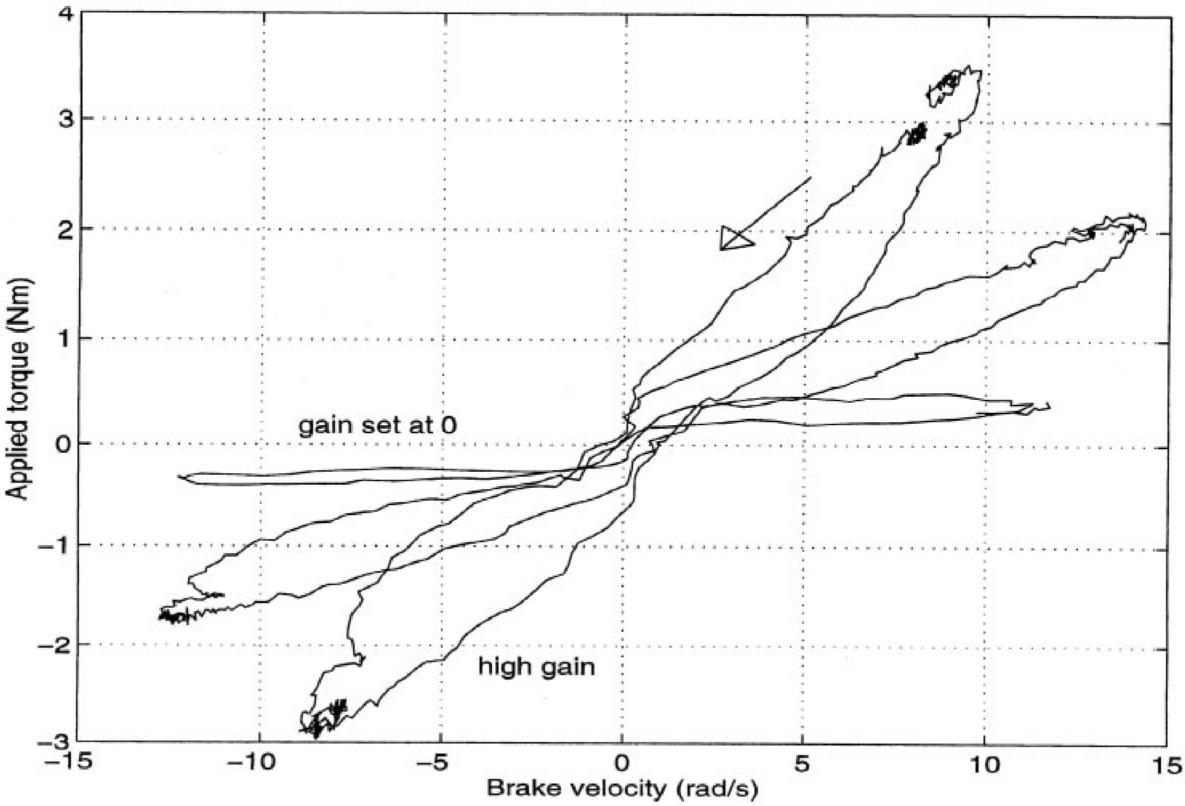


Fig. 2. B15 Particle brake based damping element tested at three gain levels

where $V_{in}$ and $i$ are the voltage and current applied to a permanent magnet D.C. motor having a torque constant (and a generator constant) of $K_T$, and an armature resistance and inductance of $R$ and $L$ respectively. If the output shaft angle $\theta$ is measured and used as the input to a P-D controller of the form $-V_{in} = P(\theta - \theta_{ref}) + D\theta$ where $\theta_{ref}$ is a desired neutral position for the impedance element, the resulting system can be tuned to have a variable impedance. Assuming $T_{in}$ to be the independent variable and $\theta_{ref}$ to be an initial condition, a general form of the transfer function for the system admittance is

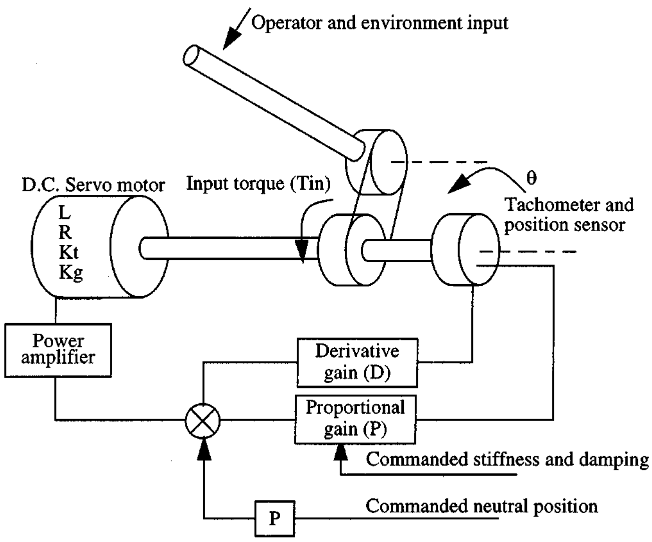


Fig. 3. Mechanical arrangement for a variable stiffness impedance

$$\theta = \frac{(R+sL)T_{in} - PK_T\theta_{ref}}{s^3JL + s^2(JR+LB) + s(RB+K^2\,T+DK_T) + PK_T}$$

From this equation the apparent stiffness of the mechanism (once transient effects can be neglected) is

$$K = \frac{PK_T}{R}$$

Thus the mechanism can establish a range of stiffnesses by adjusting $P$ as needed. Further using the Routh-Hurwitz stability criteria the following inequality must hold for stability.

$$D + K_T + \frac{RB}{K_T} - \frac{JLP}{JR+LB} > 0$$

Which, if the mechanical system has no damping, reduces to

$$D > \frac{LP}{R} - K_T$$

Thus a P-D controller can be used to set a range of system stiffnesses limited by the torque capabilities of the motor. The D term of the controller can be used to provide additional damping, but in any case must be set so that the system is stable. Where there is no mechanical damping the system inertia has no effect on stability however it does effect the period of oscillations and the settling time of the system.

**3.2.2 Results.** The linear system described above was simulated using the Mathworks Simulink program. Parameters for the simulation were based on a servo-motor mechanism used in the Cybernetics Department of the University of Reading for undergraduate teaching. This system uses the G9M4T 18 volt permanent magnet motor from Printed Motors Limited, Bordon, England. The motor has an estimated torque constant of 0.028 Nm/A and a maximum torque of 0.14 Nm. Motor inductance was estimated as 7 mH and motor resistance as 1.1 Ohms. The motor includes an integrated tachometer and a servo potentiometer and drives a load via a 5:1 belt reduction. The external damping was assumed to be negligible and two inertial loads were simulated, a $10^{-4}$ $Kgm^2$ load, representing the inertia of the motor and transmission plus a 0.1 kilogram beam 0.2 metres long; and $4x10^{-5}$ $Kgm^2$ load, representing the inertia of just the motor and transmission. Using these parameters the maximum gain using the Routh-Hurwitz stability criteria is 4.24 and is independent of mechanism inertia.

Figure 4 shows the simulated response of the servo-motor mechanism under three conditions of the P-D controller for the same torque input. The input consisted of a series of increments in torque with a final torque of 1.8 Nm. When high stiffness is required the P-D mechanical impedance exhibits a high overshoot and a lightly damped second order response. As lower impedance values are set in the P-D controller the element follows the input torque with increasingly spring like characteristics.

A comparison to the simulated results was made on the servo-motor mechanism used for undergraduate teaching in the Cybernetics Department. Mechanism stiffness was measured using both digital and analogue controllers and the range of achievable stiffness is shown in Figure 5. The overshoot characteristics seen in the simulation were not apparent in the servo-motor mechanism due to the high level of friction and damping in the mechanism.

## 4. DISCUSSION

### *4.1 Design criteria*

One primary problem of programmable impedance elements is the dissipation of heat. Magnetic particle brakes are designed to dissipate heat to a limited extent however the expectation is that this will be over the short period of time taken to halt a rotating machine. When running the brakes as a damping mechanism, the maximum power that must be dissipated occurs when the damping coefficient is set high and the brakes are opposing torque to their maximum extent. For the brake detailed in Figure 2 this condition represents a dissipation of 35 watts, significantly over the 20 watt continuous dissipation given in the specifications. Heat dissipation is similarly a factor for an impedance element based on servo-motors. The extreme operating condition occurs when the element is simulating a spring with a force applied at the limit of the stall torque of the electric motor. Under this circumstance the heat is dissipated by the resistance of the motor windings and, unlike the spring it attempts to emulate, heat must be dissipated according to an $I^2R$ relationship.

The hysteresis in the damping curve for the magnetic particle brakes shown in Figure 2 will also limit the effectiveness of these devices as programmable damping elements. At high gains the considerable increase in

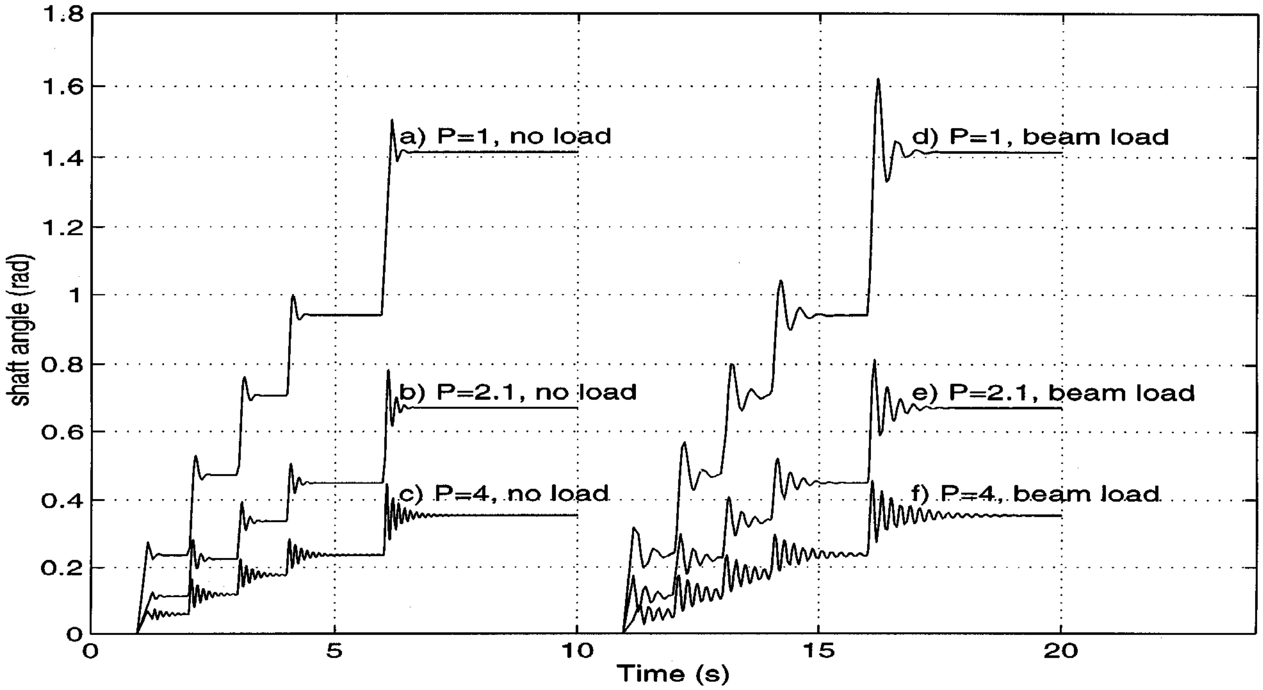


Fig. 4. Simulation of sucessive steps in input torque for three settings of the P–D controller. All curves have D = 0.01 and successive step inputs in torque producing a final torque of 1.8Nm. a) P = 1, $J = 4 \times 10^{-5}$ $Kgm^2$, b) P = 2.1, $J = 4 \times 10^{-5}$ $Kgm^2$, c)P = 4, $J = 4 \times 10^{-5}$ $Kgm^2$, d) P = 1, $J = 10^{-4}$ $Kgm^2$, e) P = 2.1, $J = 10^{-4}$ $Kgm^2$, f) P = 4, $J = 10^{-4}$ $Kgm^2$

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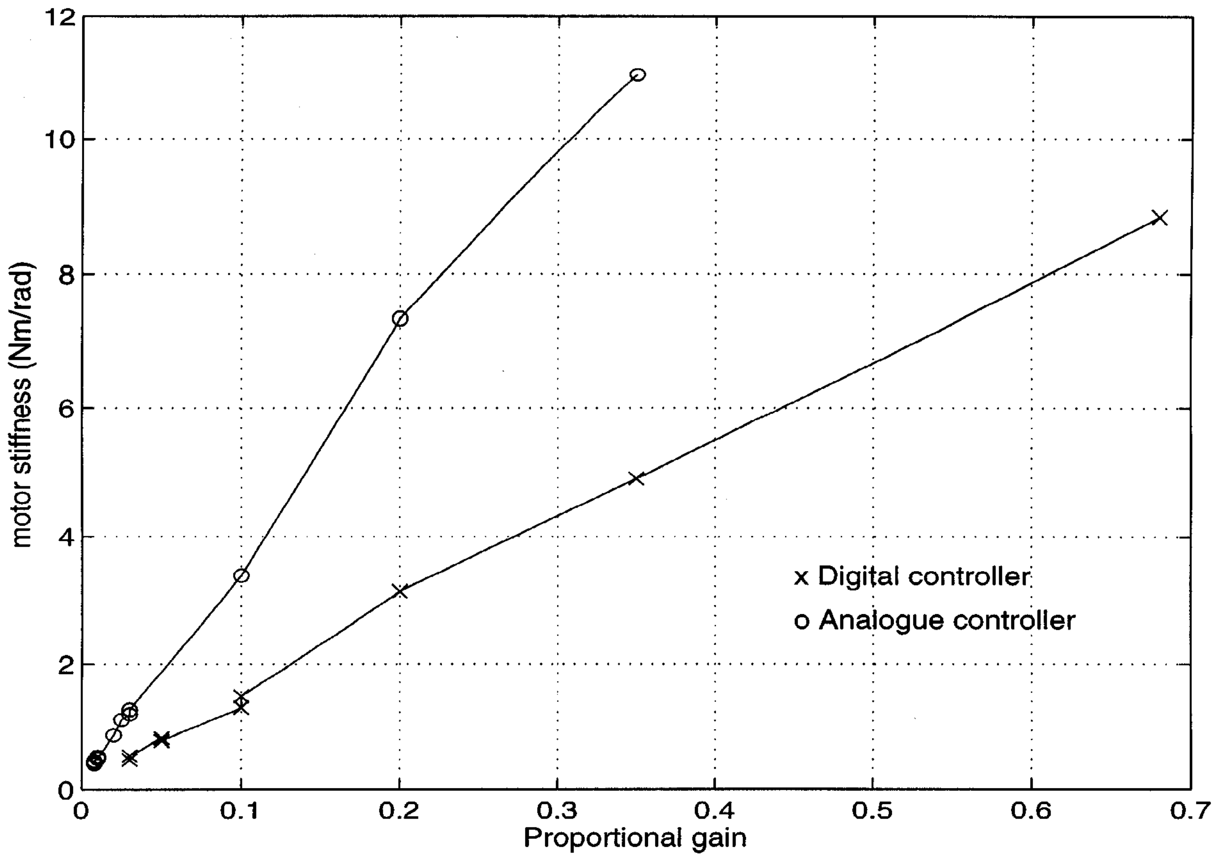


Fig. 5. Range of achievable stiffnesses using a G9M4T servo motor with a 5:1 belt reduction

hysteresis can be attributed to the building up and breaking down of the particle chains that span the distance from the brake casing to the rotating plate. In addition to the hysteresis there is a nonlinear characteristic as the damping element velocity approaches zero. The control circuit to generate programmable damping makes an assumption that braking torque is proportional to current and this relationship does not hold at low values of current. Although some of this nonlinear effect can be compensated for in the design of the brake current amplifier, most of this nonlinearity is due to friction. To achieve a more accurate programmable damping mechanism will require measurement of shaft torque. Building this measurement into an outer servo loop will reduce both the hysteresis and the nonlinearity about the origin of a programmable damping element.

The implementation of an impedance element with a detuned P-D controller shows more promise although the high level of friction in the experimental system resulted in a high level of damping away from the spring origin and a discontinuity close to the origin. The friction also had the more positive effect of increasing the margins of stability. However in most practical systems it will be necessary to minimise external damping through good mechanical design. As with the brake it will also be possible to improve performance by measuring the applied operator torque but force and torque measurements are notoriously susceptible to drift.

Because of the diverse nature of applications in rehabilitation and clinical technologies a range of impedance elements will be needed. Although conceptually similar to any off the shelf actuator, the ultimate impedance experienced by the user is strongly dependent on the mechanical linkage and the nature and position of the sensors.

### *4.2 Possible clinical applications*

Once a programmable impedance element is incorporated into a rehabilitation or clinical device there is a wide range of areas where it can be seen to have an advantage. At the lowest level an impedance element would provide a convenient way for a user or a health-care professional to adapt the mechanism to differing conditions. An example of such an application would be a mechanism for tremor suppression that can be set to different levels depending on the level of tremor. Thus a person with Parkinson's disease might use the mechanism with a light damping just after taking a tremor suppressing drug, but would set the mechanism to a high level of damping once the effects of the drug had diminished. A person with muscular dystrophy could use a programmable impedance to assist with arm movements. An arm support mechanism would superimpose a range of mechanical characteristics on the user depending on the task that was being attempted. This could be achieved in several ways. The simplest would be to establish arbitrary points in the user's region of operation and specify the controller parameters so that the manipulandum is attracted towards the nearest. These points and the apparent stiffness of the mechanism could be set by the user, or by a computer.

Where the individual has sufficient strength to move against the set impedance they could simply move from one valley of attraction to the next, with the controlling computer switching impedance parameters as predetermined boundaries are reached. If the person has lost strength to do this, the mechanism would adapt easily to a strategy where he or she used an alternate input mechanism such as voice commands, or a sip-puff switch to cycle between attraction points or to move attraction points to regions near a position of interest for the person. The use of programmable impedance has an additional advantage in that the rehabilitation mechanism will also adapt to any forces imposed by the environment and the individual. When a task required stiff movements, such as opening a desk drawer, the individual could grasp the handle and set a high stiffness in the direction of drawer movement, but a low stiffness perpendicular to drawer movement. Such an impedance field would be highly tolerant of the variations in the environment.

Thus programmable mechanical impedance elements may have a value in the design of clinical devices, as they can be adapted to the individual, can be programmed to the circumstance and could be used as an upgrade component. In contrast to mechanisms based on position servo-control where the controller is designed to achieve an end point within well defined constraints such as rise time and overshoot, a programmable impedance element is better suited to the problems of matching the mechanism to the time varying impedance properties of humans interacting with ill defined environments. In contrast a programmable damping element has the attraction of only consuming mechanical power thus giving it some attractive safety properties. Design of orthotic mechanisms based on programmable damping elements will not only allow the mechanism to suppress tremors, but can also be programmed to consider the condition and intention of the user, and the task that he or she may be trying to perform. Thus both programmable impedance elements and programmable damping elements provide an useful solution to the problems of specifying, designing and building clinical devices for rehabilitation, therapy, posture support and mechanisms to facilitate manipulation. The criteria for stability presented in this paper assumes a linear system and makes no allowance for the time for the system to settle. Humans are inherently nonlinear mechanisms and a human perception of system stability may be dramatically different from the theoretical definitions. If this stability criteria is used in a mechanism design it can only provide a conservative boundary for stability, and would only apply to a lumped system consisting of the human plus mechanism inertia, stiffness and damping. This theory primarily provides an insight for system design and further work is needed to collect data and develop reasonable human simulation models that can validate system stability. However, the perception of system stability and the person's ability to 'control' movement into and out of stable regions may be a more relevant design criteria for impedance mechanisms. This too needs further research.

One of the attractions of programmable impedance and damping elements is that they can emulate piece-wise nonlinear properties although the time taken to switch control parameters will limit the degree to which this can be achieved. Further work is needed to develop considerations for system stability since the programmable impedance makes the assumption that the human is a linear system and although there are models for the dynamic characteristics of human joints, these are only effective for free space movements.

## 5. CONCLUSION

This paper has proposed an approach to designing clinical and rehabilitation equipment based on programmable mechanical impedance elements. Two elements are suggested, a completely passive element based on particle brakes and an impedance element based on a servo-motor mechanism with a detuned P-D controller. The limits of operation of such elements and several potential applications have been identified. Stiffness and damping is a concept that is readily understood by the potential client group using this type of technology and the associated health-care professionals. If the impedance element is designed so that the results from any external programming result in a stable system under the conditions imposed by the human and environment, then it will be acceptable to allow these parameters to be adjusted to the task in hand or the clinical needs of the individual. Only the linear conditions that determine stability are given in this paper, further work will be needed to explore the stability considerations under nonlinear conditions, such as considerations of the time taken to switch P-D parameters in the controller in comparison to changes in the external environment. However a person's perception of stability may be a more relevant design metric for programmable impedance elements. A controllable damping element has none of these associated stability problems but must be designed to maintain linear properties over as wide a range of programmable damping. Such a programmable damping element is suitable for integration into mechanisms that control tremor, but for other clinical conditions it will be necessary to set the stiffness and damping of a programmable impedance element, thus raising questions about system stability and oscillations.

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