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Modeling and Preliminary Testing Socket-Residual Limb Interface Stiffness of Above-Elbow Prostheses

Jonathon W. Sensinger and
Mahidol University, Bangkok 10400, Thailand

Richard F. ff. Weir[Member, IEEE]

Biomechatronics Development Laboratory, Rehabilitation Institute of Chicago, Chicago, IL 60611 USA, and with the Jesse Brown VA Medical Center, Chicago, IL 60611 USA, and also with the Departments of Physical Medicine and Rehabilitation and Biomedical Engineering, Northwestern University, Evanston, IL 60208 USA

Jonathon W. Sensinger: jsensinger@gmail.com; Richard F. ff. Weir: rweir@northwestern.edu

Abstract

The interface between the socket and residual limb can have a significant effect on the performance of a prosthesis. Specifically, knowledge of the rotational stiffness of the socket-residual limb (S-RL) interface is extremely useful in designing new prostheses and evaluating new control paradigms, as well as in comparing existing and new socket technologies. No previous studies, however, have examined the rotational stiffness of S-RL interfaces. To address this problem, a math model is compared to a more complex finite element analysis, to see if the math model sufficiently captures the main effects of S-RL interface rotational stiffness. Both of these models are then compared to preliminary empirical testing, in which a series of X-rays, called fluoroscopy, is taken to obtain the movement of the bone relative to the socket. Force data are simultaneously recorded, and the combination of force and movement data are used to calculate the empirical rotational stiffness of elbow S-RL interface. The empirical rotational stiffness values are then compared to the models, to see if values of Young's modulus obtained in other studies at localized points may be used to determine the global rotational stiffness of the S-RL interface. Findings include agreement between the models and empirical results and the ability of persons to significantly modulate the rotational stiffness of their S-RL interface a little less than one order of magnitude. The floor and ceiling of this range depend significantly on socket length and co-contraction levels, but not on residual limb diameter or bone diameter. Measured trans-humeral S-RL interface rotational stiffness values ranged from 24–140 Nm/rad for the four subjects tested in this study.

Index Terms

Finite element analysis (FEA); fluoroscopy; joint stiffness; socket compliance

I. Introduction

The performance of prosthetic components depends on the interface between the prosthesis and the user, termed a S-RL interface and illustrated in Fig. 1. For example, in the presence of a perturbation or static load, a stiff prosthetic elbow will still be displaced if the prosthetic

elbow is in series with a compliant S-RL interface. Many biomimetic control paradigms used in prostheses [1]–[4] assume a rigid S-RL interface. If the S-RL interface is compliant, or the S-RL interface stiffness may be modulated by the user, such control paradigms become superfluous. These examples illustrate the need that engineers have for knowledge of the rotational stiffness ($k = T/\theta$) of the S-RL interface, to ensure that they do not over-design prostheses at the expense of increased cost, complexity, and development time. Specifically, what is the magnitude of a S-RL interface's rotational stiffness, and how much may it be modulated by the subject by co-contracting their muscles?

Clinical fitting of subjects and socket design are also improved by properly understanding the effect of variables that significantly influence S-RL interface stiffness. Questions such as what effect does the amputation length have on stiffness, does myodesis improve stiffness, how large of an effect does creating a distal window in the socket have on stiffness, and what effect does the conformity of the fit have on the prosthesis function, may be answered if an accurate model of the S-RL interface stiffness is available.

The rotational stiffness of the S-RL interface is a function of the material properties and anthropomorphic parameters of the residual limb. Accordingly, the first portion of this paper will develop mathematical and finite element models of the S-RL interface and analyze the sensitivity of the models to each of the variables. The second portion of the paper will empirically measure the rotational stiffness of the S-RL interface for four subjects with a transhumeral amputation, using fluoroscopy (X-ray movie frames) to measure bone movement relative to the socket. These results are compared to the models, using X-ray slides to obtain accurate anthropomorphic measurements of the parameters.

II. Modeling

Modeling of the S-RL interface rotational stiffness significantly depends on the value used for Young's modulus (E). Young's modulus is a measure of the ratio of stress to strain, for small strains. Numerous studies have characterized Young's modulus for soft tissue using mechanical indentation tests and ultrasound. Zheng *et al.* [5] present a good review of many of the studies that specifically examine residual limb soft tissue. Although several studies [6], [7] have shown tissue deformation to be nonlinear, the variance in linear Young's modulus due to co-contraction [8] and across different studies is larger than any recorded nonlinear effects. Mak *et al.* [9] have shown that for the lower limb there is no difference in Young's modulus between the amputated limb and the intact limb. They have shown that there is a large difference, however, depending on age and gender. Older subjects (57–78 years old in the study) had 40%–45% less stiffness than younger subjects (25–35 years old in the study) and could only change their Young's modulus by 24%, whereas younger subjects could change their Young's modulus by 43% [9], presumably because the older subjects could not co-contract their muscles as much as the younger subjects. Male subjects in the study had a mean modulus 40% higher than female subjects [10]. Krouskoup *et al.* [8] have demonstrated that subjects with a transradial (below elbow) amputation can modulate their Young's modulus from 6 to 100 kPa by cocontracting their muscles. In a similar study of able bodied subjects, Zheng [6] found a Young's modulus range of 10–60 kPa on the forearm. No studies have been done of transhumeral (above-elbow) residual limb soft tissue, but studies of the Young's modulus for forearm [6], [8] are in the same range as studies for the Young's modulus of thigh tissue [8], [10]. As a result, this range seems applicable to trans-humeral soft tissue in the lack of findings specific to the upper-arm region.

It may be thought that myodesis, in which residual muscle is attached to the residual bone, is required to allow for modulation of the stiffness. This does not appear to be true, however. It is unlikely that all of the subjects in Krouskoup's study [8] had myodesis due to

conventional surgical practice, yet a Young's modulus range was found of 6–100 kPa. Zheng *et al.* [11] have been able to achieve pattern recognition results using the morphological shape of trans-radial residual limb tissue, indicating that the muscle tenses, rather than merely shifting. Abboudi [12], using a similar pressure dependent recognition system, have measured the ability of subjects without myodesis to change the pressure exerted by their muscles from 0–55 kPa. As a result, it seems likely that all subjects can modulate the stiffness of their residual limb.

Another parameter that influences the stiffness of the S-RL interface is the compressibility of the tissue, defined by Poisson's ratio: $\nu = -\epsilon_{\text{transverse}}/\epsilon_{\text{axial}}$, where $\epsilon_{\text{transverse}}$ is the strain in the transverse direction and ϵ_{axial} is the strain in the axial direction. Poisson's ratio ranges from 0 to 0.5, where 0 indicates that a deformation in the axial direction has no effect on the transverse direction, and thus the volume has changed. A Poisson's ratio of 0.5 indicates that that total volume of the shape has not been affected by the deformation in one dimension. Poisson's ratio for soft tissue is assumed to range between 0.45 and 0.5 (incompressible) by many investigators, as reviewed by Zheng *et al.* [5], due to the nearly incompressible nature of tissue in general. Recent experiments have suggested ways to easily obtain Poisson's ratio *in vivo* by using two different-sized indenters [13], but have yet to apply these techniques specifically to the skin.

Both the math model and finite element analysis (FEA) model used a cylindrical geometry with a domed distal end. Anthropomorphic data, including the radius of the bone, radius of the residual limb, length of the bone in contact with the residual limb, and amount of soft tissue distal to the bone, were recorded from X-ray slides of each patient.

A. Mathematical Modeling

The stiffness of an object may be represented as a function of the area, depth, and Young's modulus of the object. The force exerted by a discrete volume within the object is equal to the stiffness, multiplied by the amount of deformation. The force may be represented by the following equation:

1) Tissue Directly in Front of Bone—

$$F = \frac{EA\Delta d}{d} \quad (1)$$

where F is the force exerted by the tissue against the socket, E is Young's modulus, A is the area under consideration, Δd is the deformation of tissue, and d is the original depth of the tissue to the bone, as illustrated in Fig. 2.

To calculate the force exerted at a specific point, A may be separated as the product of length l and bone width w_b , and the force may be differentiated with respect to length

$$dF = \frac{Ew_b\Delta d}{d} dl. \quad (2)$$

d and Δd are given by the following:

$$d = \frac{w_t - w_b}{2\cos\theta} \quad (3)$$

$$\Delta d = l \tan \theta \quad (4)$$

where θ is the angular movement of the bone relative to the socket and w_t is the thickness of the tissue. Substituting 3 and 4 into 2, and simplifying

$$dF = \frac{2Ew_b l \sin \theta}{w_t - w_b} dl. \quad (5)$$

The moment at point l is obtained by multiplying dF by l . The majority of the moment across l acts in one direction. There is a small portion, however, that acts in the opposite direction. This transition occurs when

$$l = l' = \frac{w_t (1 - \cos \theta)}{2 \sin \theta}. \quad (6)$$

The sum of the moments above and below the center of rotation is

$$M = 2 \left(\int_0^{S/2} \frac{2Ew_b l^2 \sin \theta}{w_t - w_b} dl - \int_0^{l'} \frac{2Ew_b l^2 \sin \theta}{w_t - w_b} dl \right) \quad (7)$$

where S is the length of the humerus within the socket. When integrated, this equation becomes

$$M = -Ew_b \left(\frac{w_t^3 \cos^2 \theta - \cos \theta \sin \theta S^3 - 2w_t^3 \cos \theta + w_t^3 - \sin \theta S^3}{6(w_t - w_b)(\cos \theta + 1)} \right). \quad (8)$$

By ignoring the portion of the moment that acts in the opposite direction (of which both the displacements and moment arm are small, producing small torques) and simplifying the trigonometry to act near $\theta = 0$, the equation simplifies to a format that may easily be displayed in stiffness form as

$$K = E \frac{w_b S^3}{6(w_t - w_b)}. \quad (9)$$

2) Tissue Connected to Deformed Tissue—The tissue directly in front of the bone will deform by the same amount that the bone moves relative to the socket. As a result, (4) adequately captures the resulting force caused by the tissue directly in front of the bone, because the amount of deformation is known. The amount of deformation of the surrounding tissue, however, is unknown. The resulting deformation will be greatest near the edge of the bone, and minimal near the skin. The deformation, illustrated in Fig. 3, will be determined by that geometry which minimizes the stored energy, where energy is a function both of the compressive and shear forces resulting from the displaced tissue.

Biaxial compressive forces and shear forces are applicable to this shape, for which

$$dU_{\text{compression}} = \frac{E}{2(1-\nu^2)} (\varepsilon_x^2 + \varepsilon_y^2 - 2\nu\varepsilon_x\varepsilon_y) dV \quad (10)$$

and

$$dU_{\text{shear}} = \frac{G\gamma^2}{2} dV \quad (11)$$

where G is the shear modulus, and equals $E/2(1+\nu)$, ε_x and ε_y are the strains, dV is the discrete volume being analyzed, and γ is the angle of shear deformation. An exponential curve was found to best minimize the sum of energy over the entire residual limb. The power of the exponent depends on the subject's anthropomorphic parameters and the amount of deformation at the boundary of the bone. As a result, the equation cannot be algebraically solved, but may be solved using discrete algebraic code. The results of this mathematical equation for a 0.1 rad (6°) rotation are given in the results section.

B. Finite Element Modeling

A finite element model of the S-RL interface was created using Pro/Mechanica Wildfire 2.0 (Parametric Technology Corporation, Needham, MA). Static analysis of tetrahedra solid P elements using multipass adaptive (MPA) element fitting was performed for all analyses. MPA analysis records the maximum error at element intersections in Von Mises stress (distortion energy) and strain energy error, increasing the order of the polynomials used to fit the geometry until the errors are all less than 10%. The amount of acceptable error may be lowered, but lowering this limit results in a dramatic increase in computational time, since the boundary of contact interface is complex. The setup is illustrated in Fig. 4.

Fixed contact points were created at equally spaced intervals along the interface. The tissue that interfaces with the bone was fixed, and a 18 N force was applied to the end of the prosthesis to create a 5 Nm torque, which is realistic of the loads encountered by amputees. Analyses were done using values of $E = 10$ kPa and 70 kPa, and $\nu = 0.4, 0.5$, and ~ 0 (0.0001). The range of Poisson's ratio was chosen to represent a slightly larger range of the variable (0.4–0.5) than commonly assumed in other literature (0.45–0.5), and to assess the sensitivity of the model to the variable across its entire range of physical possibility by modeling it at either extreme (0.0001–0.5). The results are shown in Table II.

III. Fluoroscopy

To empirically measure the rotational stiffness of S-RL interfaces, the movement of the humerus relative to the socket was measured in four subjects using fluoroscopy: a type of low-dosage X-ray. The protocol was approved by the Northwestern University Institutional Review Board, and all patients signed informed consent forms.

A. Subjects

Four subjects, including three males and one female, were included in this study. Socket type, age, and anthropomorphic data are given in Table I. Graphical representations of their residual limb are given in Fig. 5.

B. Apparatus and Procedures

Fluoroscopy slides (a rapid series of low-radiation X-rays) were recorded at 5–8 Hz using a Philips Easy Diagnost Super 80CP digital RF system. Lead markers were taped to the socket to accurately align the socket in X-ray slides. Force data were collected using an Omega LCFA-50 (Omega, Stanford, CT) tension/compression load cell and recorded in Matlab (Mathworks, Natick, MA) at 40 Hz. The author manually applied a sinusoidal load on the load cell at the distal portion of their prosthesis. The average magnitude was ± 7 Nm at 0.5 Hz. An illustration of a slide is given in Fig. 6.

Subjects leaned over the X-ray table, abducting their shoulder to allow the arm to lie parallel to the floor. The first subject's elbow was locked in 45° of flexion (135° between the forearm and arm). The remaining three subject's prosthetic elbows were locked in 90° of flexion. The experimental setup for this study did not permit the examination of co-contraction modulation by subjects. Because the viewing window of the fluoroscopy machine was small (13 cm diameter), subjects had to maintain the position of their residual limb in the presence of force perturbations. As a result, their muscles were always moderately co-contracted to maintain this position.

C. Data Analysis

An edge detecting filter was applied to a cropped version of each slide using Matlab's *edge* function with a canny filter. The humerus was then manually aligned in each slide. The angular movement of the markers with respect to the aligned humerus was recorded. This method preserved any rotational movement, while compensating for any sliding movement. Otherwise, sliding movement would influence the recorded rotational movement.

The inertial force on the load cell from the prosthesis was calculated based on the inertia of the prosthesis (0.4 kg m^2) and the calculated rotational acceleration. Rotational inertia was calculated by measuring the frequency at which the prosthesis swung as a pendulum. Rotational acceleration, due to the sparse nature of position data points, was obtained by integrating the torque measurements to obtain position, and then fitting the scaling and shifting coefficients until the integrated position matched the actual position. The inertial component of forces was subtracted from the force data, and the remaining force was compared to the change in angular position. Due to the low frequency of oscillation, the inertial load was small for all subjects (0.01%–10%).

Because there is noise present in the marker measurements obtained from the X-ray slides, rotational stiffness may either be found by using a linear least squares regression of the compliance, and then inverting the slope to obtain rotational stiffness, or by calculating the first principal component using principal component analysis—both techniques yielded the same result.

IV. Results

A. Math Model

Sensitivity studies were done using the discrete math model to see how significantly it diverged from the simple math model, which did not consider the effect of surrounding tissue. Each of the anthropomorphic variables were individually varied for each subject, over a range larger than that encountered across all four subjects. Bone and residual limb diameter did not have large effects on the resulting rotational stiffness. The socket length, however, did have a significant effect, and the rotational stiffness may accurately be represented across subjects as the following function:

$$K=[0.13 - 0.15]ES^{[2.7-2.75]} \quad (12)$$

where E is Young's Modulus and S is the socket length.

B. Math Model Compared With FEA Model

The math model and FEA model are compared with each other in Table II for subject A, using several values for Poisson's ratio and Young's modulus. These sensitivity studies were only done for a single subject since each analysis took a substantial amount of time to complete.

From Table II it may be seen that in the math model, rotational stiffness is proportional to Young's modulus for the math model, and almost proportional to Young's modulus for the FEA model. The range of Poisson's ratio encountered in soft tissue (0.4–0.5) has a small effect on the rotational stiffness. The math model corresponds well to the FEA model.

C. Empirical Results

The measured stiffness ranged from 25–138 Nm/rad across the four subjects. High individual linear correlation ($r^2 = 0.84 - 0.97$) was obtained, indicating that a linear model provided a good fit for the residual-limb socket interface stiffness. The results from a sample subject are shown in Fig. 7, and the mathematical FEA and empirical rotational stiffness for each subject are tabulated in Table III.

V. Discussion

Rotational stiffness of the S-RL interface is proportional to Young's modulus and significantly smaller than the rotational stiffness of conventional prosthetic elbows. As a result, even in the presence of a stiff prosthetic elbow the overall rotational stiffness of the elbow and S-RL interface may be controlled by co-contraction of the subject. Because of this inherent ability of subjects to modulate the stiffness of their S-RL interface, it appears that control paradigms that implement impedance control directly distal to the S-RL interface are superfluous. These paradigms are still useful, however, on joints that are distally removed from the S-RL interface. For example, the stiffness of the wrist joint is unaffected by the stiffness of a trans-humeral S-RL interface. Likewise, the stiffness of an elbow joint is unaffected by the stiffness of a S-RL interface at higher amputation levels, such as shoulder disarticulation. As a result, impedance control still merits investigation.

Whether Poisson's ratio is 0.4 (nearly incompressible) or 0.5 (completely incompressible) does not have a large effect on the rotational stiffness of the S-RL interface compared to the change in rotational stiffness that results as Young's modulus changes with co-contraction. Because the models correspond well with the empirical measurements, local estimations of Young's modulus may be applied to a global S-RL interface. This finding should extend the results found in this paper to other joints, for which Young's modulus is locally known, but where global models have not yet been made.

Pro/Mechanical Wildfire is unable to perform large-deformation FEA on models with contact regions. As a result, a small-deformation contact model was used. This is a limitation of the study, and future studies should use software that is capable of performing large-deformation FEA on contact regions. For the purposes of this preliminary study, however, the model sufficiently correlates with the empirical evidence to be useful for design of future prostheses.

Likewise, a linear Young's modulus was used in this analysis, whereas in reality soft-tissue has a nonlinear Young's modulus. The range of linear Young's modulus reported in the literature is so large as to minimize the significance of any reported nonlinearity, however. Until a consensus is reached in the literature on the nonlinear properties of the residual limb, a simple linear model provides a sufficiently accurate model, and significantly simplifies any analysis.

Ideally, individual Young's modulus ranges would have been obtained for the four subjects tested in this study. In this experiment, empirical results were collected first, and a model was created second, and as a result, no thought was given to observing individual stiffness values. The first two subjects used a standard harness to suspend their socket. The latter two subjects used a silicone liner to suspend their socket. A silicone liner typically compresses the soft tissue more than a hard socket, in turn providing a stiffer residual limb due to precompression of the soft tissue. This effect cannot be extracted from the data however, due to the small sample size.

A. Clinical Implications

The discrete math model accurately captures the main effects of the FE model, with the exception of Poisson's ratio. As a result, the adjusted math model (11) provides a simple model of the effect of anthropomorphic factors on the S-RL interface rotational stiffness. This equation may be used to evaluate prosthetic components, control algorithms, and socket techniques.

For example, the S-RL interface rotational stiffness significantly increases as the socket length increases. The effect becomes more dramatic as the socket length increases. As a result, creating a distal window in the socket may relieve discomfort without affecting the subject's ability to provide torque. It does, however, decrease the stiffness of the S-RL interface by 4–20 Nm/rad for every 1 cm of window, depending on the original length of the socket. A distal window reduces the stiffness more for long sockets than for short sockets. Conversely, salvaging an extra centimeter of distal bone increases the stiffness of the S-RL interface by 4–20 Nm/rad. Likewise, proximally extending the socket increases the stiffness by 4–20 Nm/rad.

As another clinical example, the math models indicate that tissue medial and lateral to the bone have a large effect on the stiffness of the socket. As a result, creating large medial and lateral windows in the socket will have an impact on the rotational stiffness of the S-RL interface. Creating small windows, however, will not affect the stiffness, as long as they are limited to the sagittal plane.

VI. Conclusion

The measured S-RL interface stiffness of four subjects is within the modeled range of stiffness predicted using math and FE models, using Young's moduli available in the literature. This same technique may be applied to other joints. From these models, it appears that persons can modulate the rotational stiffness of their S-RL interface over a wide range of values. The floor and ceiling of this range depend significantly on socket length and co-contraction levels, but not on residual limb diameter or bone diameter. Measured transhumeral S-RL interface rotational stiffness values ranged from 24–140 Nm/rad for the four subjects tested in this study. Control paradigms that modulate the stiffness of a joint are unnecessary when directly in series with this S-RL interface, but may still prove to be useful when implemented in more distal joints or when the level of amputation is more proximal.

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References

1. Sensinger, JW.; Weir, RFF. Impedance control using a compact series elastic actuator. presented at the IEEE/ASME Int. Conf. Mechatronic Embedded Syst. Appl; Beijing, China. 2006.
2. Sensinger, JW.; Weir, RFF. Non-backdrivable impedance control. presented at the Robotics: Sci. Syst.; Robotic Syst. Rehabil., Exoskeleton, Prosthetics Workshop; Philadelphia, PA. 2006.
3. Abul-Haj CJ, Hogan N. Functional assessment of control-systems for cybernetic elbow prostheses—Part 2: Application of the technique. *IEEE Trans Biomed Eng.* Nov; 1990 37(11):1037–1047. [PubMed: 2276751]
4. Popat RA, Krebs DE, Mansfield J, Russell D, Clancy E, Gillbody KM, Hogan N. Quantitative assessment of 4 men using above-elbow prosthetic control. *Arc Phys Med Rehabil.* 1993; 74:720–729.
5. Zheng Y, Mak AFT, Leung AK. State-of-the-art methods for geometric and biomechanical assessments of residual limbs: A review. *J Rehabil Res Develop.* 2001; 38:487–504.
6. Zheng YP, Mak AFT, Lue B. Objective assessment of limb tissue elasticity: Development of a manual indentation procedure. *J Rehabil Res Develop.* 1999; 36:71–85.
7. Silver-Thorn MB. In vivo indentation of lower extremity limb soft tissues. *IEEE Trans Rehabil Eng.* Sep; 1999 7(3):268–277. [PubMed: 10498373]
8. Krouskop TA, Dougherty DR, Vinson FS. A pulsed doppler ultrasonic system for making noninvasive measurements of the mechanical properties of soft tissues. *J Rehabil Res.* 1987; 24:1–8.
9. Mak AFT, Liu GHW, Lee SY. Biomechanical assessment of below-knee residual limb tissue. *J Rehabil Res Develop.* 1994; 31:188–198.
10. Zheng Y, Mak AFT. Effective elastic properties for lower limb soft tissues from manual indentation experiment. *IEEE Trans Rehabil Eng.* Sep; 1999 7(3):188–198.
11. Zheng Y, Chan MMF, Shi J, Chen X, Huang QH. Sono-myography: Monitoring morphological changes of forearm muscles in actions with the feasibility for the control of powered prosthesis. *Med Eng Phys.* 2005; 28(5):405–415. [PubMed: 16115790]
12. Abboudi RL, Glass CA, Newby NA, Flint JA, Craelius W. A biomimetic controller for a multifinger prosthesis. *IEEE Trans Rehabil Eng.* 1999; 7(2):121–129. [PubMed: 10391581]
13. Choi APC, Zheng YP. Estimation of Young's modulus and poisson's ratio of soft tissue from indentation using two different-sized indentors: Finite element analysis of the finite deformation effect. *Med Biol Eng Comput.* 2005; 43:258–264. [PubMed: 15865137]

Biographies

Jonathon W. Sensinger received the B.S. degree in bioengineering from the University of Illinois, Chicago, in 2002, and the M.S. and Ph.D. degrees in biomedical engineering from Northwestern University, Evanston, IL, in 2005 and 2007, respectively.

He worked as a Research Scientist at the Rehabilitation Institute of Chicago in 2007 in biomechatronics, pattern recognition, and haptics research under the direction of Dr. T. Kuiken. He is currently a Professor in the Biomedical Engineering Department of Mahidol University in Bangkok, Thailand.

Richard F. ff. Weir (M'91) was born in Dublin, Ireland. He received the B.A. degree in mathematics and the B.A.I. in microelectronics and electrical engineering from Trinity College, Dublin, Ireland, in 1983, and the M.S. and Ph.D. degrees in biomedical engineering from Northwestern University, Evanston, IL.

He is currently a Senior Research Scientist by the Rehabilitation Institute of Chicago, Chicago, IL, and holds appointments as a Research Scientist at the Jesse Brown VA Medical Center, Chicago, IL, and as a Research Associate Professor in the Departments of Physical Medicine and Rehabilitation and Biomedical Engineering, Northwestern University. He has research interests in the areas of neural engineering, biomechatronics and rehabilitation, specifically, artificial arm/hand systems, manipulators, robotics, and their associated control. The current focus of this work is the development of a multichannel/multifunction prosthetic hand/arm controller based on implanted myoelectric sensors (IMES), the development of an externally-powered partial hand prostheses, and the development of multifunction externally-powered prosthetic hand. The work presented here is part of a new initiative to explore the utility of series elastic actuators in nonbackdrivable drive systems for prosthetics applications.

Fig. 1. S-RL interface compliance between socket and residual limb. The interface is not rigid: some compliance is present (illustrated here by a pair of springs), and affects control of the prosthesis.

Fig. 2. Deformation of tissue when a load is applied to the socket. (a) Undeformed S-RL interface: b is the diameter of the bone; f is the diameter of the soft tissue; f is the length of the socket. (b) Deformed S-RL interface.

Fig. 3.

Transverse plane depiction of tissue deformation. Deformation at d_1 is known, because the bone determines the amount of deformation. Deformation at d_2 , however, is unknown. Some deformation will occur because the tissue is connected to the tissue directly in front of the bone. The amount of deformation may be calculated by finding that shape which minimizes the total energy stored at that segment.

Fig. 4. (A) FEA p -mesh of CAD geometry. (B) Shape of the deformed tissue inside of the socket for a 2.8 Nm load, where $E = 70$ kPa and $\nu = 0.4$.

Fig. 5. Graphical representation of subject residual limb. Anthropomorphic parameters were measured from X-ray slides of the residual limb.

Fig. 6.
(a) Fluoroscopy slide of S-RL interface. (b) Edges obtained on a cropped version using Matlab's edge command.

Fig. 7.
Empirical rotational stiffness of S-RL interface.

TABLE I

Subject Anthropomorphic Data. All Measurements are in centimeters

<i>Subject (age)</i>	Socket Type	Bone Dia.	Limb Dia.	Distance from Elbow to:	
				distal end of bone	proximal end of socket
A (27)	Harness	2.3	8.5	19	33
B (39)	Harness	2.9	10.3	19	35
C (43)	Suction	2.4	10.0	17	34
D (50)	Suction	1.7	10.0	9	29

All measurements are in cm

TABLE II

Results of Math and FEA Analysis for a Sample Subject

		Modeled Young's Modulus for Subject A					
		E=10 kPa			E=70 kPa		
		Math	FEA	Error	Math	FEA	Error
Poisson's Ratio	$\nu=0$	6.9	N/A	N/A	K=48.6	K=41.3	18%
	$\nu=0.4$	K=5.9	K=6.8	13%	K=41.5	K=48.8	15%
	$\nu=0.5$	K=5.8	K=7.8	26%	K=40.8	N/A	N/A

K is in units of Nm/rad. These results were calculated for an applied torque of 2.8 Nm

TABLE III

Modeled and Measured S-RL Interface Rotational Stiffness for Four Subjects

Subject	Stiffness (Nm/rad)				
	Math relaxed – cocontracted	FEM relaxed –cocontracted	Measured		
			Average	95% Confidence range	r ²
A	4–59	4–70	25	23–28	0.87
B	6–103	6–100	138	96–244	0.84
C	6–99	5–83	38	32–45	0.85
D	9–156	14–227	81	78–85	0.97