

Correlation between active and passive isometric force and intramuscular pressure in the isolated rabbit tibialis anterior muscle

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Abstract

The purpose of this study was to quantify the relationship between intramuscular pressure (IMP) and muscle force during isometric muscle contraction of the rabbit tibialis anterior (TA) absent the effect of either bone or fascia. To quantify this relationship, length–tension experiments were performed on the isolated TA of the New Zealand White rabbit (mass = 2.5 ± 0.5 kg, $n = 12$). The knee was fixed in a custom jig, the distal tendon of the TA was attached to a servomotor, and a 360 μm fiber optic pressure transducer was inserted into the TA. The peroneal nerve was stimulated to define optimal length (L_0). The length–tension curve was created using 40 Hz isometric contractions with 2-min rest intervals between each contraction. Measurements began at $L_0 - 50\%L_f$ and progressed to $L_0 + 50\%L_f$, changing the length–tension in 5% L_f increments after each contraction. Qualitatively, the length–tension curve for isometric contractions was mimicked by the length–pressure curve for both active and passive conditions. Linear regression was performed individually for each animal for the ascending and descending limb of the length–tension curve and for active and passive conditions. Pressure–force coefficients of determination ranged from 0.138–0.963 for the active ascending limb and 0.343–0.947 for the active descending limb. Passive pressure coefficients of determination ranged from 0.045–0.842 for the ascending limb and 0.672–0.982 for the descending limb. These data indicate that IMP measurement provide a fairly accurate index of relative muscle force, especially at muscle lengths longer than optimal.

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1. Introduction

It is very difficult to study the mechanical properties of individual skeletal muscles during normal movement. This is not only due to the technical difficulties associated with invasive measurements but also because human locomotion results from coordinated interaction among numerous skeletal muscles, tendons, and joints. Studying muscle during movement is further complicated by the fact that significant muscle force may result from active muscle contractions or due to simply to passive muscle length changes.

In a few rare cases, muscle force has been measured directly during normal locomotion both in cats (Gregor

et al., 1988; Walmsley et al., 1978; Walmsley and Proske, 1981) and in humans (Gregor et al., 1991; Komi et al., 1992). Measurement of muscle forces and activation patterns during locomotion under various conditions has provided a general paradigm of motor control across muscles of differing architectures and fiber. In human studies, buckle transducers were acutely implanted on the Achilles tendon and used to measure force over a range of dynamic movements. While this approach yields powerful information, it is highly invasive and impossible to apply to the clinical setting. Furthermore, the buckle transducer used in humans only measures total tension of the Achilles complex rather than the individual tension contributions of the soleus and gastrocnemius muscles. Recently, a fiber optic-based transducer was used to obtain tendon forces in the Achilles complex and patellar tendon of humans (Finni et al., 2000; Komi et al., 1996). While the method

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provides a valuable research tool that is much less invasive compared to the buckle transducer, it is also inappropriate for clinical application.

Over two decades ago, [Baumann et al. \(1979\)](#) suggested that intramuscular pressure (IMP) could provide an estimate of muscle tension. The precise mechanism for the generation of IMP is debatable; however, IMP has been defined as the hydrostatic fluid pressure within a muscle ([Sejersted et al., 1984](#)). While previous investigators have measured IMP during isometric torque ([Sejersted and Hargens, 1995](#); [Sejersted et al., 1984](#)), eccentric and concentric joint contraction ([Aratow et al., 1993](#); [Ballard et al., 1998](#); [Crenshaw et al., 1995](#)), as well as during walking and running ([Ballard et al., 1998](#)), these investigators assumed that IMP reflects muscle force. This is because there has never been a direct comparison between IMP and muscle force. Further, the relative contribution of active and passive muscle force to IMP has not been studied and remains unknown. Such knowledge is a prerequisite to using IMP as a meaningful tool.

Based on the promise that IMP could provide an estimate of active and passive muscle tension, the purpose of this study was to correlate intramuscular pressure with direct and simultaneous measurement of muscle force during active and passive conditions. The hypothesis examined by this study was that intramuscular pressure is a measurable mechanical parameter that is directly related to two independent phenomena—passive elongation of muscle fibers and active force generation by muscle fibers.

2. Methods

The experimental model used was the tibialis anterior (TA) muscle of the New Zealand White rabbit (mass = $2.5 \text{ kg} \pm 0.5$). This model was selected primarily based on the accessibility of the TA, its 3° pennation angle and thus parallel fiber arrangement ([Lieber and Blevins, 1989](#)), and predominantly fast fiber type percentage, believed to be representative of other limb muscles ([Aigner et al., 1993](#); [Peter et al., 1972](#)). Based on 12 subject's average TA physiologic cross-sectional area (PCSA) of 65 mm^2 and the transducer cross-sectional area of 0.10 mm^2 , the transducer represented about 0.2% of the muscle PCSA resulting in minimal direct muscle trauma. Control experiments demonstrated that neither insertion of the transducer nor the number of contractions affected isometric force after forty of the 40 Hz isometric contractions imposed under the conditions described here. Animal preparation and measurement of isometric contractile properties were performed essentially as previously described ([Lieber and Fridén, 1993](#); [Lieber et al., 1991](#)). The protocol was approved by the University of California, San Diego, San Diego

State University, and the VA Medical Center committees on the Use of Animal Subjects in Research. All experimental procedures adhered to the guidelines set forth by the National Institute of Health "Guide for the Care and Use of Animals."

Briefly, rabbits were induced on 4% and maintained on 2% halothane (21/min). Heart rate and oxygen saturation were monitored (VetOx™, Heska Co., Fort Collins, CO) throughout the test duration and anesthesia was adjusted as needed. A midline incision was made from the ankle to the mid-section of the thigh. The leg was immobilized using 3.2 mm Steinmann pins placed in the mid-tibial and distal femoral condyles and secured to a custom jig. The distal biceps femoris was released and the peroneal nerve exposed. A cuff electrode was placed around the peroneal nerve for direct muscle activation (Pulsar 6Bp Stimulator FHC Inc., Bowdoinham, ME). The TA fascia was completely removed, the distal tendon transected, and the elevated muscle attached to a servomotor (Cambridge Model 310B, Aurora Scientific Inc., Ont., Canada) and aligned with the force-generating axis of the motor ([Fig. 1](#)). This physical arrangement was chosen so that only the muscular contributions to IMP would be studied and the confounding influence of fascia ([Garfin et al., 1981](#); [McDermott et al., 1982](#)) and bone ([Gershuni et al., 1984](#)) would be avoided. The motor was calibrated by hanging known masses from the motor arm and measuring their respective voltages (Calibration factor 2.0 V/kg , $r = 0.997$, $p < 0.0001$). Muscle temperature was maintained at 37°C with radiant heat, mineral oil, and a servo-temperature controller (Model 73A, YSI, Yellow Springs, OH). A $360 \mu\text{m}$ diameter fiber optic pressure sensor (Luna Innovations Inc., Blacksburg, VA) was inserted via an 18-gauge angiocatheter at approximately 10° in line with the force-generating axis of the fibers and at the thickest proximal portion of the muscle ([Fig. 1](#)). The pressure transducer output was adjusted to zero volts immediately after insertion into the muscle. The pressure transducer's performance characteristics show a mean accuracy of $1.45 \pm 0.32\%$ and a mean repeatability of $1.5 \pm 0.81\%$ ($n = 5$ different transducers tested in vitro) the details of which are described elsewhere ([Kaufman et al., 2003](#)).

Stimulation threshold voltage was determined by increasing the current delivered to the peroneal nerve until peak force was reached. Supramaximal voltage was then calculated by doubling the threshold voltage. All contractions were performed at supramaximal voltage to ensure maximal activation of all TA motor units. Twitch optimal length (L_0) and peak force (P_0) were determined and tension was measured during twitch and tetanic contractions at stimulation frequencies of 5, 10, 20, 40, 60, 80, and 100 Hz and a pulse width of 0.3 ms.

The length-tension protocol consisted of 40 Hz contraction trains over a 600 ms period with a 2-min

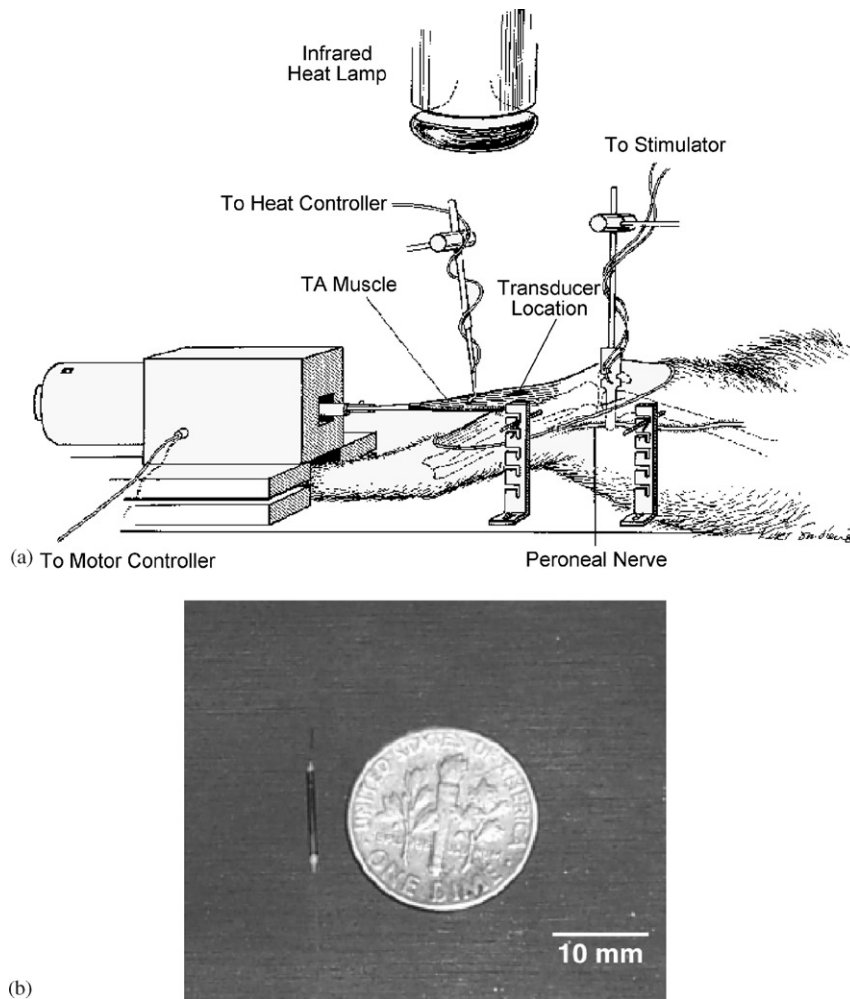


Fig. 1. Experimental apparatus used to measure isometric force–intramuscular pressure relationship. (A) Rabbit hindlimb immobilized in custom jig via Steinmann pins securing the distal femur and proximal tibia. (B) Photograph of pressure transducer used relative to the size of a dime. Scale bar = 10 mm.

rest interval interposed between each contraction. Muscle fiber length (L_f) was calculated from muscle length using the rabbit TA fiber length-to-muscle length ratio (Lieber and Bleivins, 1989). Measurements began at a slack length, $L_0 - 50\%L_f$ and were increased in 5% L_f increments until a length of $L_0 + 50\%L_f$ was reached. Passive muscle force was defined as the resting muscle force at each length and measured for each contraction bout during the 100 ms time period prior to muscle stimulation. Length, tension, pressure, and temperature were recorded for each contraction using a data acquisition board (610E series, National Instruments, Austin, TX) and a LabView virtual instrument (National Instruments, Austin, TX) acquiring data at 4 kHz.

Active and passive tension records were converted to stress for the purpose of reporting force as a function of stimulation frequency and calculating RMS error by dividing tension by the muscle's calculated PCSA.

The muscle's PCSA was calculated using the equation described by Sacks and Roy (Sacks and Roy, 1982). In order to estimate the "goodness of fit" between the force and pressure data, linear regression was performed for each animal subject for the ascending and descending limbs of the active and passive relationships. The ascending limb was defined as lengths less than or equal to L_0 , while the descending limb, was defined as lengths greater than L_0 . In order to understand the underlying contributions to the IMP signal, the data were separated into the ascending and descending limbs. The length–tension relationship was arbitrarily divided into these two regions based on the fundamentally different subcellular events occurring in both regions of the curve (Gordon et al., 1966). In this way, we hoped to understand the mechanism of IMP generation as well as quantify the correlation with isometric force. RMS error was also calculated to provide an estimate of the variance between stress

and IMP for each subject across the entire length–tension relationship. The calculation used for RMS error was

$$\text{RMS error} = \sqrt{\sum_{i=1}^{20} \frac{(\sigma_{m_i} - Pm_i)^2}{n}}$$

where σ_{m_i} represents muscle stress at the i th length, Pm_i represents intramuscular pressure at the i th length, and n represents the number of contractions ($n = 20$).

3. Results

In general, raw muscle isometric force–time traces had the same appearance as the raw IMP–time traces. During the initial experiment, as muscle stimulation frequency increased, IMP also increased (Fig. 2). Correlation between isometric force and IMP was very high as a function of stimulation frequency (average $r^2 = 0.95$, average $p < 0.01$). Unfortunately, due to the relatively slow frequency response of the microsensor instrumentation, it was not possible to determine the precise temporal relationship between isometric force and IMP. As expected, the fiber length–isometric tension curve was characterized by an “ascending limb” at lengths less than L_0 and a “descending limb” at lengths greater than L_0 (Fig. 3A). The shape of this curve presumably represents a scaled and distorted version of the sarcomere length–tension curve previously published (Gordon et al., 1966; Rack and Westbury, 1969). Passive muscle tension increased in a nearly exponential fashion at lengths beyond optimal. The length–pressure relationship generally mimicked the shape of the length–tension curve with an ascending limb at lengths less than L_0 and descending limb at lengths greater than L_0 (Fig. 3B). However, it was clear, based on the large standard errors and irregular form of the curve, that pressure proved more variable compared to tension. Interestingly, at lengths above 35% L_f passive tension began to decrease, presumably due to injury of passive muscle structures such as the surrounding connective tissue or intracellular parallel structures. This drop was also reflected in the passive pressure curve at corresponding lengths.

A positive linear relationship was found between IMP and force on both the ascending and descending limbs. Table 1 summarizes the results of the linear regression analysis. Pressure–force coefficients of determination (i.e., r^2) ranged from 0.138–0.963 ($p = 0.0001$ –0.3221) for the active ascending limb and 0.343–0.947 ($p = 0.0001$ –0.0024) for the active descending limb. Passive pressure coefficients of determination ranged from 0.045–0.842 ($p = 0.0005$ –0.5581) for the ascending limb and 0.672–0.982 ($p = 0.0001$ –0.002) for the

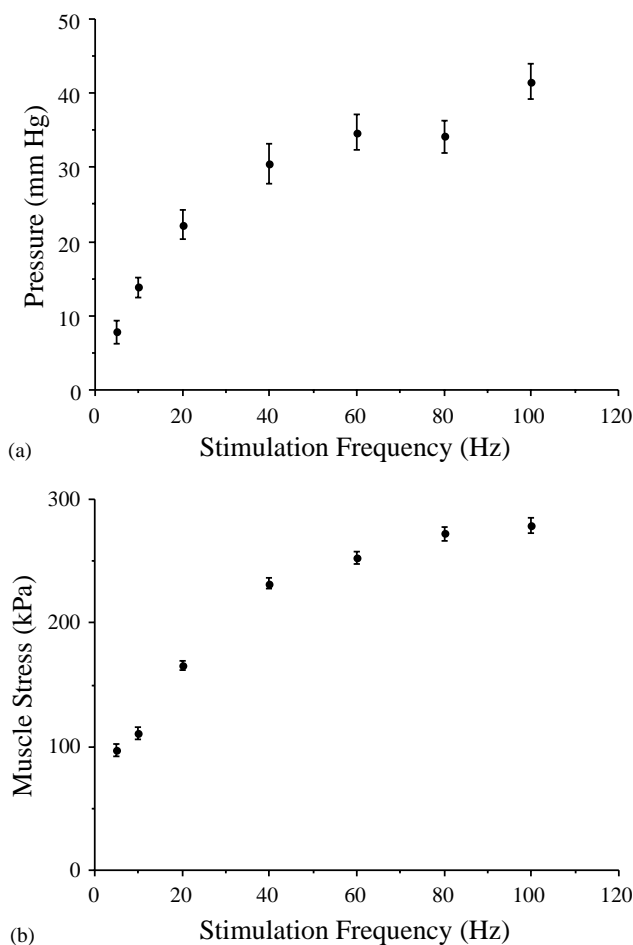


Fig. 2. Relationship between isometric stress and intramuscular pressure for the rabbit tibialis anterior during stimulation at frequencies ranging from 5 to 100 Hz. Stress and pressure were well-correlated across this range of frequencies with the linear regression relationship being y (IMP in mm Hg) = $0.014x$ (Isometric stress in kPa), ($p < 0.01$, average $r^2 = 0.64 + 0.28$, $n = 10$). Data are plotted as mean \pm SEM.

descending limb. Correlations were higher for the descending limb of the length–tension curve compared to the ascending limb for both active and passive conditions, while activation did not affect the goodness of fit statistic (Table 1).

Individual normalized active stress and IMP values were plotted for lengths ranging from $L_0 - 50\%L_f$ to $L_0 + 50\%L_f$. RMS error was used to quantify the variance between individual active length–IMP and their respective length–stress curves. The values for the individual variance between active IMP and stress across all lengths tested is expressed in Table 1, and ranged from 76.5–239 kPa. Passive records imitated the individual passive length–stress curves, but had a higher degree of variability compared to the active curves. Individual RMS errors were also calculated between passive IMP and stress. Passive variance values are expressed in Table 1 and ranged from 9.49–45.0 kPa. In

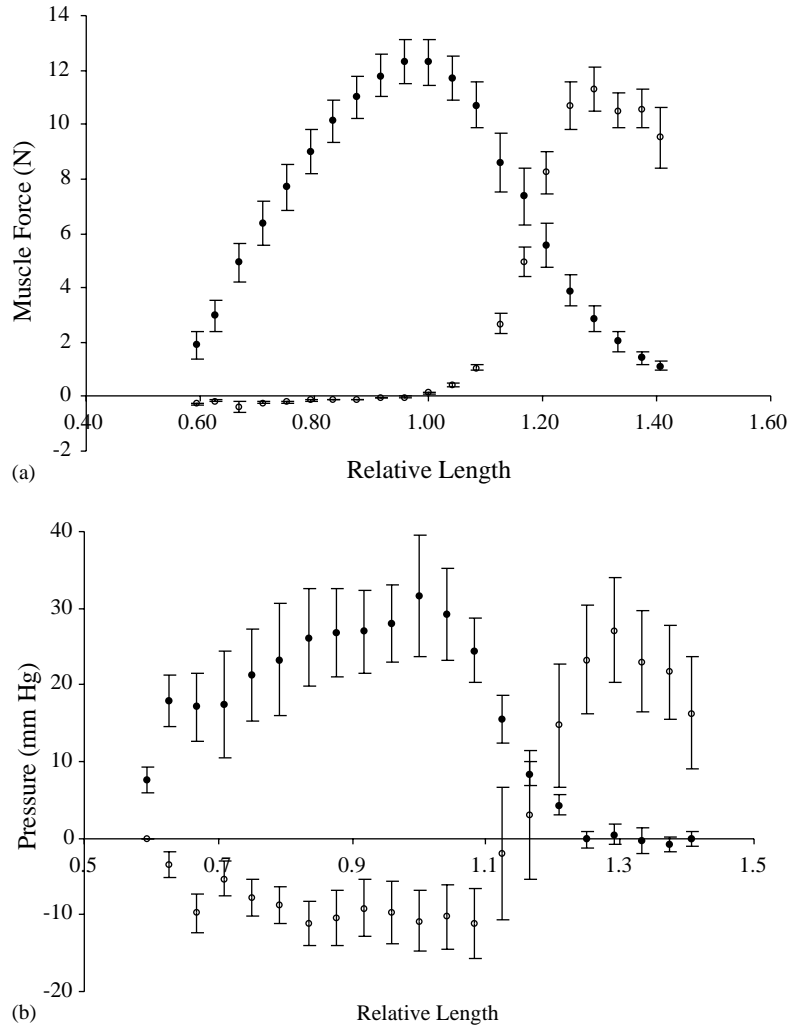


Fig. 3. Relationship between relative muscle length and (A) isometric force or (B) intramuscular pressure for the rabbit tibialis anterior. Filled symbols represent measurements from activated muscles while open symbols represent measurement from passive muscles. Force and pressure were better correlated at long lengths, independent of activation state (see Table 1 for details). Data are plotted as mean \pm SEM.

Table 1
Summary of isometric stress–pressure correlations^a

Activation state	Limb of the length–tension curve	RMS error (kPa)	Pressure coefficients of determination (r^2)	
			Mean \pm SEM	Range (min–max)
Active	Ascending limb	175.2 \pm 52.98	0.60 \pm 0.25	0.138–0.963
	Descending limb	147.19 \pm 47.46	0.77 \pm 0.16	0.343–0.947
Passive	Ascending limb	10.18 \pm 4.21	0.53 \pm 0.36	0.045–0.842
	Descending limb	20.19 \pm 13.83	0.86 \pm 0.11	0.672–0.982

^a Values represent mean \pm standard error for $n = 10$ animal subjects.

this case, the lower numbers do not reflect a better correlation, rather they represent smaller absolute stress and pressure values achieved during passive compared to active contraction.

4. Discussion

The purpose of this study was to quantify directly the relationship between IMP and isometric muscle force

across a range of muscle lengths in rabbit tibialis anterior muscle under both active and passive conditions. Previous investigators have implied (Aratow et al., 1993; Ballard et al., 1998; Korner et al., 1984; Sadamoto et al., 1983; Sejersted et al., 1984) that muscle tension and IMP were correlated but the direct relationship between isolated muscle force and IMP had never been measured. If muscle pressure and force are well correlated, IMP may be used to estimate muscle force.

The results of this investigation demonstrated that the correlation between isometric force and IMP was high as a function of stimulation frequency ($r^2 = 0.95$) and that active and passive IMP mimicked active and passive muscle force across a range of lengths ($L_0 + 50\%L_f$ to $L_0 - 50\%L_f$). Linear regression at lengths less than L_0 (ascending limb) and greater than L_0 (descending limb) for both active and passive conditions demonstrated that the correlations were slightly higher on the descending limb compared to the ascending limb and that muscle activation alone did not dramatically alter these correlations.

The average length–pressure curve mimicked the average length–tension curve. The length–tension curve was highly reproducible with variability similar to that measured in other studies (average coefficient of variation across subjects = 47%). The length–pressure curve was more variable, particularly at lengths less than L_0 (average coefficient of variation across subjects = 96%). This variability was manifest as differences between individual pressure coefficients of determination for the active ascending and descending limb. Regression analysis revealed that, on average, 60% of the variability between active isometric muscle force and IMP was explained for lengths less than L_0 , but 77% of the variability was explained at lengths greater than L_0 . A linear correlation between IMP and isometric torque in human muscle has been shown in previous literature with similar variability between subjects (Baumann et al., 1979; Sadamoto et al., 1983; Sejersted et al., 1984). In previous studies, this variability was attributed to differences in fiber geometry, muscle thickness, and depth of placement of the transducer within the muscle. While these factors may contribute to differences between subjects, we suggest that the increased variability on the ascending limb may be attributed to movement of the transducer between fibers, since we obtained approximately the same degree of variability between subjects in this inbred rabbit strain. We suggest that, as a muscle contracts, fibers expand and compress fluid that ultimately registers a pressure in the transducer head. In one rare case, we observed that the transducer was forced out of the muscle when it was not inserted far enough and the muscle was highly shortened. Perhaps smaller such movements occurred during other contractions, although this was not directly measured. The basis for the difference in correlation

coefficients between the ascending and descending limb is not known; however, we suggest that, at longer lengths (on the descending limb) lateral compression between fibers secured the transducer in a fixed position enabling pressure on the transducer head. At slack lengths, the sensor was not as well secured and was able to move within the muscle when the fibers expanded during contraction.

In considering the use IMP as an indirect index of muscle force, the ability to measure passive muscle properties is imperative. As shown in Figs. 3A and B the passive length–IMP curve mimicked the passive length–tension curve. Similar to active IMP, the descending limb correlation between passive IMP and muscle forces was better than the ascending limb. Specifically, 86% of the variability between passive muscle force and IMP was explained on the descending limb while only 53% of the variability was explained on the ascending limb. Unlike passive tension, which remained close to zero from $L_0 - 50\%L_f$ to L_0 , negative passive pressures were recorded under these conditions. Previous investigations have reported negative relaxation pressures (Ballard et al., 1998; Crenshaw et al., 1992), which were thought to result from measurement error such as movement of the catheter during contraction or placement of the sensor near bone. Ballard et al. suggested that negative pressures could be a physiologic response that is necessary for muscle perfusion when contracting. The basis for these negative pressures remains unknown, but an alternative hypothesis may be that at slack lengths the transducer can move between the muscle fibers and that a small vacuum is created around the transducer as the fibers pull away from it. As length increases, fibers may become tighter around sensor reducing the amount the muscle fibers can pull away from the transducer. Despite negative pressures it is encouraging that a passive IMP had a significant correlation with muscle force.

Another intriguing finding was the pressure response to lengths greater than $L_0 + 35\%L_f$ (Fig. 3B). Like force, at these lengths pressure peaked and continued to decline. It is believed that the decline in passive force at lengths greater than $L_0 + 35\%L_f$ is due to passive injury. Interestingly, IMP appeared to be sensitive to the passive injury.

The reason for variability between subject's pressure coefficients of determination for all conditions was further explored. Several authors have implicated transducer depth, intact compartments, and varying muscle architecture as causes for variation (Aratow et al., 1993; Jarvholm et al., 1991; Sejersted et al., 1984). The aforementioned factors should affect local IMP, if IMP is theoretically determined by muscle thickness, fiber radius, and fiber stress as described (Skalak, 1982). It is not reasonable, however, to think that these factors are the sole cause of intersubject variability. Another

factor may be fluid flow. To determine if a fluid response was a factor, a pilot experiment was conducted ($n = 3$). Each experiment consisted of a 30 s contraction at 40 Hz in a maximally stimulated TA muscle, which represents a nearly fused contraction in rabbit TA. Tension peaked and then slowly decayed over the 30 s until stimulation ceased and tension declined to zero. IMP was collected simultaneously and mimicked tension (Fig. 4). The tension IMP response was consistent across all animals with no oscillations seen in the pressure signal. We therefore suggest that, over time IMP, was responding directly to muscle tension rather than indirectly to fluid flow.

In summary, IMP accurately predicts muscle force under this study's conditions namely: longitudinal muscle fiber architecture and with the muscle isolated and measured isometrically. Potential limitations of this study that were not rigorously controlled were the difficulty of reliably placing the transducer in the same muscle location and anchoring the transducer into the muscle. The model that we used was an isolated muscle, which may behave differently compared to muscle in an intact compartment. Fascia was removed to determine

the isolated effect of muscle on IMP. We do not have explicit information regarding the effect of fascial removal but suspect, based on the literature, that absolute pressures in the absence of fascia would be lower. This is speculative, however, since fascial compliance will decrease with increasing muscle length, perhaps causing a loss in linearity of the force–IMP relationship. The effect of fascia is beyond the scope of the present study but has been addressed in previous investigations (Garfin et al., 1981; McDermott et al., 1982; Pedowitz et al., 1990; Styf and Korner, 1986). In addition, we have no information regarding the way in which muscle isometric force scales quantitatively with IMP. Finally, for IMP to be useful as an in vivo muscle force estimating tool, the relationship between IMP and dynamic muscle force must also be determined. Future studies are required to quantify many of these other sources of variability.

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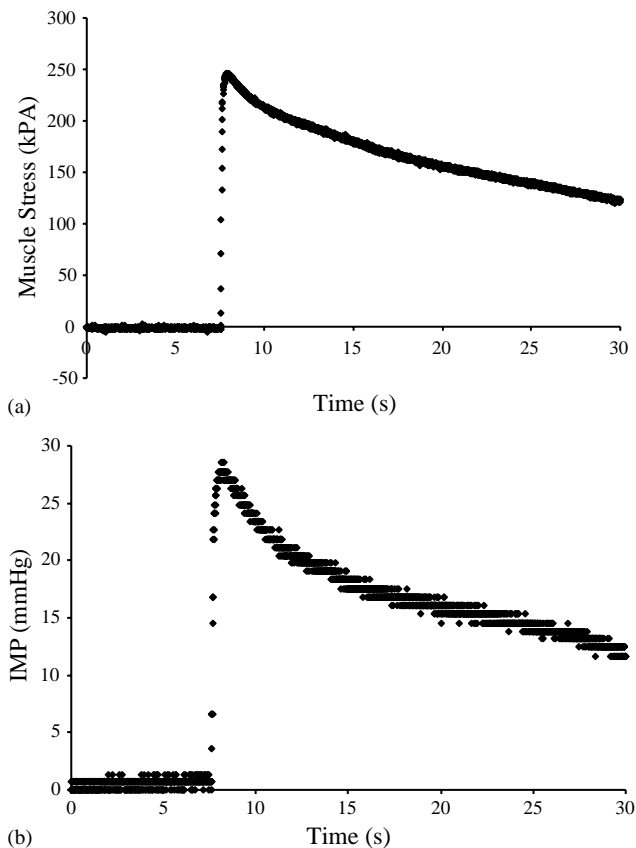


Fig. 4. Time course of (A) isometric muscle stress or (B) intramuscular pressure during a prolonged tetanic contraction at a stimulation frequency of 40 Hz. Note the parallel changes in stress and IMP suggesting that IMP is directly affected by muscle stress rather than indirectly by secondary fluid flow.

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