Mechanical behavior of the quadriceps femoris muscle tendon unit during low-load contractions

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Finni T, Havu M, Sinha S, Usenius J-P, Cheng S. Mechanical behavior of the quadriceps femoris muscle tendon unit during low-load contractions. J Appl Physiol 104: 1320–1328, 2008. First published March 13, 2008; doi:10.1152/japplphysiol.01069.2007.—We examined the relationships between morphology and muscle-tendon dynamics of the quadriceps femoris muscle of 11 men using velocity-encoded phase-contrast magnetic resonance imaging (MRI). Thigh muscle electromyography and joint range of motion were first measured outside the MRI scanner during knee extension-flexion tasks that were performed at a rate of 40 times/min with elastic bands providing peak resistance of 5.2 kp (SD 0.4) to the extension. The same movement was repeated inside the MRI scanner bore where tissue velocities and muscle morphology were recorded. The average displacement in the proximal and distal halves of the rectus femoris and vastus intermedius aponeuoses was different (P = 0.049), reflecting shortening (1.6%), but the tensile strain along the length of the aponeuoses was uniform. The aponeuosis behavior varied among individuals, and these individual patterns were best explained by the differences in relative cross-sectional area of rectus femoris to vastus muscles (r = 0.71, P = 0.014). During dynamic contraction, considerable deformation of muscles in the axial plane caused an anatomic measure such as muscle thickness to change differently (decrease or increase) in different sites of measurement. For example, when analyzed from the axial images, the vastus lateralis thickness did not change (P = 0.946) in the frontal plane through femur but increased in a 45° oblique plane between the frontal and sagittal planes (P = 0.004). The present observations of the heterogeneity and individual behavior emphasize the fact that single-point measurements do not always reflect the overall behavior of muscle-tendon unit.

TO UNDERSTAND the muscle-tendon mechanics during locomotion, it is necessary to know the detailed architecture of the muscle and its dynamics during contractions. Action of a muscle is functionally coupled with its adjacent structures, creating a complex interaction between passive and active tissues. Importantly, the force from the active muscle is transmitted longitudinally to tendons and laterally to the neighboring agonist and even antagonist muscles (15). In humans, studies on the relative movement between muscles and strain in the aponeurosis between the muscles may provide information on the force transmission (2, 8, 9, 13).

Our laboratory has previously reported the structure-function relationships in human calf muscles (8, 9, 13). In the present study, we investigated the architecture and tissue dynamics in the quadriceps femoris (QF) muscle undergoing dynamic contractions. The quadriceps femoris has two separate muscles, the two-joint rectus femoris (RF) and one-joint vastus muscles that attach via tendon to the patella, and are innervated by branches from the femoral nerve. The vastus muscles, or vasti, comprise three compartments: lateralis (VL), medialis (VM), and intermedius (VI). VI has a further derivative called articular muscle (musculus articularis genus) (21). The anatomic boundaries between VI and VL are not discrete in the proximal part but are fused for ~25–35% of their length (5). The RF muscle is typically described as a simple bipennate muscle, but detailed analyses have revealed a more complex structure. The proximal one-third of RF is unipennate while the distal muscle is radially bipennate. The fibers originate from an internal tendon and attach to the aponeurosis of insertion that surrounds the muscle belly (11). Thus RF is not ideal for two-dimensional (2-D) analysis using ultrasonography because a perpendicular view of muscle fascicles is difficult to obtain because of the orientation of the fascicles.

There is a major aponeurosis structure, the proximal extension of the quadriceps tendon, attached to RF that also extends inside the vastus muscles. Because all of the four knee extensor muscles transmit forces via this connective tissue, it is likely to have a role in normal and abnormal knee mechanics (10). Thus we were specifically interested in the dynamics and strain distribution in the connective tissue structure that resides between RF and the vasti. Based on our laboratory’s previous findings on strain heterogeneity in the soleus aponeurosis (9), we hypothesized that the strain distribution is nonuniform also in the QF aponeurosis.

In addition to examining the morphology and dynamics of QF muscles, we tested new methodological approaches of the velocity-encoded cine phase-contrast (VE-PC) magnetic resonance imaging (MRI) that allow 1) analysis to be faster for clinical purposes and 2) 2-D analysis of tissue movement from unidirectionally encoded VE-PC magnetic resonance (MR) images.

METHODS

Ethical Approval

Written informed consent was obtained in writing from the subjects before the measurements, and the study was conducted according to the Declaration of Helsinki. Approval for the project was obtained before its initiation from the Ethics Committee of the Central Hospital of Central Finland.

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Subjects

Eleven male subjects [age 26 yr (SD 3), height 177 cm (SD 6), weight 72 kg (SD 10)] volunteered to the study. Background information regarding physical activity and medical history was collected via a questionnaire after the informed consent was obtained. All the subjects were healthy and had no history of injuries in lower legs that could affect the results. Their physical activity background was heterogeneous; nine strength trained about three times per week while two were exercising only occasionally.

Cyclic Knee Extension

The subjects were asked to perform knee extension-flexion exercise repeatedly with their right leg while lying in a prone position. An elastic band was attached to the subject’s ankle to provide calibrated resistance to the extension (Fig. 1). When stretched, the elastic band provided a mean peak load of 5.2 kp (SD 0.4). The total work within the extension-flexion cycle was kept constant throughout the test. The mean range of knee angular displacement was 28° (SD 9) (Fig. 2). One test consisted of ~70 continuous extension-flexion movements at a rate of 40 cycles/min. The subjects followed the rhythm from a column of rising and lowering lights in front of them. The subjects were first tested in a laboratory to get them familiar with the knee extension-flexion task and to measure muscle activity [electromyography (EMG)] during the test. After that, the test was repeated in the MRI setting.

EMG

The extension-flexion exercise was first done in laboratory in the same position as in the MRI, with a goniometer attached to the knee joint and EMG electrodes attached to the thigh muscles. Muscle activity was measured from the RF, VL, VM, and biceps femoris muscles with a ME6000 biomonitor (Mega Electronics, Kuopio, Finland) using a sampling frequency of 1 kHz. The electrodes with a 20-mm interelectrode distance were placed to shaved, abraded, and cleansed sites over the muscle bellies following the recommendations of SENIAM (12). The subjects also performed maximal isometric knee extension at a knee angle of 150° [maximal voluntary contraction (MVC)]. The averaged rectified EMG (aEMG) during the active phases of the cyclic test was normalized to the subject’s MVC trial. Cross-correlation coefficients between RF-VL and RF-VM muscles were calculated.

MRI

Anatomic and functional (muscle velocity) information of the subjects’ QF muscles was gathered using a torso phase array coil in MRI (1.5-T GE Signa CV/i scanner, GE Medical Systems, Waukesha, WI). The task used in the functional tests was the same as in the laboratory setting, but it was performed without the EMG electrodes and the goniometer. To initiate the MRI data acquisition in the same movement phase in each cycle, an optical trigger device was connected to the subject’s ankle (Nomir Oy, Tampere, Finland). The task was repeated at a rate of 40 (SD 1) contractions/min. Repeatability of the contractions in MRI was ensured by setting the arrhythmia rejection rate to 10% so that the duration of the cycles had to match the preset length within a 150 ms (SD 4) margin or they were excluded from the data acquisition. The velocity of the tissue movement during the contraction cycles was acquired using VE-PC MRI (3, 9, 22). This method was used to produce two sets of 256 × 256 pixel images, each consisting of 20 temporal phases of the movement cycle, 75 ms (SD 2) apart depending on the repetition rate. One set of images contained velocity information (phase-contrast images) and the other anatomic information (magnitude images). The VE-PC scans were taken from two axial locations and from one sagittal location bisecting the RF muscle (Fig. 3). In addition, we tested one subject in supine position to evaluate the effect of posture on the muscle deformation during contraction.

The VE-PC scans were used to quantify the velocity of the muscle tissue during the extension-flexion cycle. VE-PC scans quantify the...
velocity of a coherent flow of protons by means of the phase shift they generate in the detected signal. This is routinely utilized in MR angiography to visualize blood flow in vessels and myocardium during cardiac cycle. Similar coherent motion of protons occurs in skeletal muscles during muscle contractions (1, 3, 8, 9, 20, 22). In the case of muscles, the measured velocity provides information of tissue movement during contraction and, when taken adjacent to tendon or aponeurosis, displacement and strain of those structures (see VE-PC Image Processing) (8, 9).

The VE-PC scans were acquired with a FAST GRE phase-contrast sequence, with a velocity encoding (VENC) range of ±120 mm/s, 320 mm field of view (FOV), 9.3 ms repetition time (TR), 4.6 ms echo time (TE), 30° flip angle, 5 mm slice thickness, 256 × 128 matrix, 100% phase-FOV, 1 average (NEX), and 4 views per segment (VPS) (axial sections). The parameters for sagittal VE-PC scans were VENC 120 mm/s, FOV 320 mm, TR 9.2 ms, TE 4.5 ms, flip angle 30°, slice thickness 5 mm, matrix 256 × 128, phase FOV 75%, NEX 3, and VPS 4. The spatial resolution was 1.25 mm × 2.50 mm × 5.00 mm, and the temporal resolution was 75 ms. Velocity was encoded in the cranial-to-caudal direction.

In addition to the dynamic MRI scans, typical morphological images were acquired in the axial and sagittal planes by using proton density fast spin echo (FSE) sequence. The axial images with slice spacing of 8.5 mm were used for determining muscle cross-sectional area (CSA) and volume. Muscle volumes were calculated from cross-sectional areas multiplied by distance between two slices (distance = slice spacing + slice thickness).

VE-PC Image Processing

Filtering. To filter out noise, a convolution filter with a 3 × 3 Gaussian kernel was applied to velocity images. An average velocity image was then calculated and subtracted from each frame in the phase-contrast series to remove the eddy current-induced additive phase error (3, 22).

Analysis. To analyze aponeurosis behavior, regions of interest (ROI) were placed on the images at 5-mm intervals over the RF and vastus intermedius (VI) muscle-aponeurosis interface (Fig. 3). An average of 31 ROIs per side was analyzed. Their course was followed throughout the cycle by calculating their position in each image from their previous location and the measured velocity. Because the aponeurosis is not a straight vertical line, errors to the tracking are introduced if the lateral movement is not corrected for. After calculating the position of the ROI using a velocity image, we checked its location from the next magnitude image. If necessary, the ROI placement was corrected by moving it back to the muscle-aponeurosis interface. This procedure was repeated after every image, i.e., 20 times. By placing the ROIs on a clearly distinguishable structure and using the magnitude images for determining lateral movement allows 2-D analysis of tissue movement even in cases where only unidirectional (superior-inferior direction) velocity encoding is used. This so-called pseudo-2-D analysis of the movement enables shorter scan times and higher spatial resolution than encoding the velocity in three dimensions. Furthermore, we developed an automated tracking method for the pseudo-2-D analysis, which uses a rudimentary edge
detection algorithm with Sobel filter to trace the muscle-aponeurosis interface. (See the video supplement in the online version of this article.)

Axial-to-sagittal comparison. To speed up the tracking process from the sagittal images, we tested whether the axial velocity images can be used for faster analysis in clinical work. The velocities from specific ROIs in RF and VI in the axial images were selected, and the same areas were projected to sagittal images (Fig. 3). These velocities were compared. To evaluate the effect of repeatability on our comparison between the two planes, we compared the velocities in two consecutive sagittal scans. Furthermore, we compared the tissue displacement calculated from sagittal images in two different ways: 1) tracking ROI displacement using the 20 velocity images and 2) predicting ROI displacement using the velocities from the 20 velocity images but always using the initial position of the ROI to read the velocity value (i.e., no tracking of the tissue).

Tissue displacement and strain. Tissue displacement was calculated from the velocity images by following the position of the ROIs along the aponeurosis. The displacement values were normalized to the length of the femur. To test the hypothesis of nonuniformity, the visible aponeurosis was divided into proximal and distal halves. A linear fit to the maximum displacement data was constructed 1) for the length of the entire aponeurosis and 2) as a two-part linear fit with a break at the midpoint of the aponeurosis. Tensile strains were defined as the derivatives of these linear fits. Tissue displacements and strains in the 50% proximal and distal parts of the measured area were compared. Classification of the aponeurosis behavior into three types was done based on 1) the relative movement and 2) tensile strain of the RF and VI side of the aponeurosis. When classified by the relative movement between RF and VI, type DI was defined as significantly greater maximum displacement on the RF side than on the VI side ($P < 0.05$). When classified by strain, type SI was defined as at least 25% greater strain in VI than in RF. In type SII, the strain was below 1% or differed less than 25% between the RF and VI. In type SIII, the RF side of the aponeurosis had at least 25% more strain than the VI side.

Morphological analyses. RF distal tendon length was measured from the tip of the patella to the muscle-tendon junction. The thigh length was measured from the lateral epicondyle of femur to the lateral tip of greater trochanter. Quadriceps tendon moment arm was defined as a distance between RF tendon (anterior tip of patella) and femoral intercondylar fossa at the level of lateral epicondyle. The CSA of RF and vastus muscle group were analyzed from the FSE images with the open-source software OsiriX. Because the thigh length varied between subjects, the results were normalized to 15 samples. The reproducibility of drawing the areas was high with correlation of 0.99 and mean error of 0.4% (SD 0.3) for one analyzer. The axial magnitude images from the VE-PC series were analyzed for deformation and CSA of quadriceps muscle during contraction. Two magnitude images, one from the passive phase of the cycle and one from the active phase of the cycle were chosen. The muscle outlines were drawn, and the images were superimposed onto each other with the femur aligned. VL and VI muscle thickness was measured from these images in the frontal and sagittal planes bisecting the femur and in an oblique plane at a 45° angle between the two (Fig. 4). Thickness was defined as a distance of straight line between the superficial and deep aponeurosis of the muscle.

Statistics

The variation between the velocities measured from axial and sagittal VE-PC images, and the variation between the displacements

![Fig. 4. Deformation of quadriceps femoris muscle during submaximal dynamic contraction from 1 subject. The vastus muscles and visible aponeurosis between compartments and RF muscle are outlined from the magnitude images. In the right panel, the black outline is relaxed muscle, and gray line is contracted muscle. The changes in VI and VL muscle thicknesses were measured from the locations indicated by dotted lines. The dotted vertical line represents sagittal plane, and dotted horizontal line is the frontal plane. The dotted slope represents the 45° plane of analysis. The third set of outlined images (bottom row) is from a suspended leg in supine position obtained using gradient echo sequence.](image-url)
calculated from the sagittal VE-PC images using ROI tracking or flow through a stationary ROI, were assessed by calculating typical error ($s = s_{diff}\sqrt{2}$, where $s_{diff}$ is SD of differences between 2 values (14)). Differences in displacement, strain, and EMG in different muscles (RF and vasti) were assessed using independent $t$-test after testing the equality of variances. If the variances were not equal, nonparametric Mann-Whitney $U$-test was applied. The nonuniformity hypothesis was also tested with a $t$-test by comparing the residual sum of squares of the one- and two-part linear maximum displacement models. The difference between the changes of muscle thickness was assessed by univariate analysis of variance. Spearman’s correlation was applied when relationships between types of shear and independent variables were examined. SPSS 14.0 software was used to run the analysis. The level of significance was set to $P < 0.05$.

RESULTS

Anatomy and Deformation

Example of deformation of QF muscle during contraction in prone position with leg lying flat on the surface is shown in Fig. 4. Similar deformation was observed in supine position with leg suspended. Typically, RF muscle moved laterally during the contraction. Because of the lateral muscle deformation, a ROI defined in a single axial image does not necessarily represent the same muscle tissue in the rest of the 20 images. We quantified that the minimum number of pixels common to ROIs in all images was 66% (SD 19) in RF and 91% (SD 3) in vasti, when the ROI was defined by the muscle outline.

The volume of the vastus muscle group was 1,748 cm$^3$ (SD 565), and the volume of RF was 257 cm$^3$ (SD 85). On average, the volume of RF to vasti was 18% (SD 2). The muscle volume correlated significantly with the largest CSA in both vasti ($r = 0.99, P = 0.000$) and RF ($r = 0.86, P = 0.001$). The CSAs and the relative CSA of RF to vasti are shown in Fig. 5. The CSA of RF and vasti varied along the length of the muscle. The relative area of RF was minimum (4% SD 2) in the most distal section and maximum (31% SD 6) in the proximal muscle (Fig. 5). There was no detectable difference in the CSA of the vasti between relaxed and submaximally contracted conditions when measured from the magnitude images of VE-PC sequence.

When measured from the proximal axial MR image, a significant change in muscle thickness due to contraction was observed in VI in the sagittal plane (from 1.9 to 2.2 cm; $P = 0.000$). In the proximal section the aponeuroses between the VI and VL were not always visible, so their common thickness was measured. It differed between the relaxed and contracted conditions in the 45° plane (from 4.7 to 5.1 cm; $P = 0.000$).

When measured from the distal axial MR image, a significant change in muscle thickness from relaxed to contracted condition was observed in the VL muscle at the 45° plane (from 2.0 to 2.3 cm; $P = 0.004$), in VI in the 45° plane (1.4 to 1.5 cm; $P = 0.023$) and in the sagittal plane (from 1.6 to 1.9 cm; $P = 0.002$). When the change in muscle thickness was compared between the sites of observation, there were differences in VL between the frontal and the 45° plane ($P = 0.022$, power 0.64) and in VI between all three planes ($P = 0.000$, power 0.99) (see Fig. 4).

Mean RF tendon length was 82.1 mm (SD 15.1, range 62.5–102.0 mm). The ratio of RF tendon length to thigh length was 0.18 (SD 0.03). The quadriceps tendon moment arm was 52.8 mm (SD 2.4).

Muscle Activity

Example of EMG activity levels during the knee extension-flexion exercise is shown in Fig. 2. Relative to MVC, the activity during the dynamic exercise was 9.1% (SD 8.4) in RF and 5.0% (SD 3.4) in the vastus muscles. For half of the subjects the RF muscle was predominantly activated (as in Fig. 2), whereas other subjects showed predominance of VL and VM aEMG. Peak cross-correlation coefficient and phase shift were not different between VL-RF and VM-RF muscles.

Muscle Velocity

When analyzed from the distal and proximal locations of the sagittal images (see Fig. 3), the velocity in both RF and VI was greater in the distal than in the proximal part, illustrating that the mid muscle bellies, as measured from the regions of interest shown in Fig. 3, were shortening during low-load concentric contraction. The peak muscle velocities measured from the site of the distal cross section were 41 mm/s (SD 15) in RF and 56 mm/s (SD 19) in VI (Fig. 2B).

Aponeurosis Displacement and Strain

The aponeurosis displacement was measured from the muscle just adjacent to the aponeurosis (see Fig. 3 and video supplement in the online version of this article). On the average, the VI and RF side of the aponeurosis moved the same distance during the concentric phase of the movement. Individual analysis of the maximum displacement showed that in three subjects the VI side of the aponeurosis moved more than RF (displacement type DI), in four subjects there were no differences in VI and RF displacement (type DII), and in four subjects the RF side moved more than VI (type DIII; Fig. 6). The difference in mean displacement of the distal and proximal halves of the aponeurosis was significant ($P = 0.049$), reflecting nonuniform behavior.
A derivative of the linear fit to the maximum displacement data showed that the average strain in the aponeurosis was \(-1.6\%\), meaning that the aponeurosis shortened slightly. Detailed analysis, however, revealed considerable individual variability in strain and allowed classification into three different patterns of shear in the VI-RF interface: 1) the RF side of the aponeurosis shortened more than the VI side \((n = 4)\), 2) the two shortened equally much \((n = 3)\), and 3) the VI side shortened more than the RF side \((n = 4)\) (Fig. 6). The two-part linear fit used to test the nonuniformity hypothesis showed no significant differences in strain between distal and proximal aponeurosis.

There was a correlation between the displacement-derived shear type and the relative volume of RF to vasti \((\rho = 0.71, \ P = 0.014)\) (Fig. 7), but no correlation between the strain-derived shear type and the relative volume was found.

**Methodological Testing**

The muscle velocities from the axial and sagittal VE-PC images had a correlation coefficient of 0.93 and a typical error of 7.3 mm/s (Fig. 3). For VI the correlation and typical error were 0.94 and 7.5 mm/s and for RF 0.91 and 7.2 mm/s, respectively. Velocities in two consecutive sagittal scans on a single subject had a correlation coefficient of 0.95 and a typical error of 8.9 mm/s.

The displacements calculated by tracking the ROI voxels in the sagittal VE-PC images compared with the displacements calculated from the flow through a stationary ROI had a correlation coefficient of 0.98 and a typical error of 0.7 mm. For VI the correlation and typical error were 0.98 and 0.7 mm and for RF 0.95 and 0.9 mm, respectively.

Maximum displacements were compared between the pseudo-2-D and one-dimensional (1-D) tracking methods and between automatic and semiautomatic pseudo-2-D tracking methods (Fig. 8). The correlation coefficient and typical error for pseudo-2-D and 1-D tracking were 0.92 and 0.9 mm for VI and were 0.37 and 2.5 mm for RF, respectively. The correlation coefficient and typical error for automatic and semiautomatic pseudo-2-D tracking were 0.94 and 0.9 mm for VI and were 0.92 and 1.0 mm for RF, respectively.

**DISCUSSION**

The main findings of the study were that during a low-load concentric muscle action, the quadriceps aponeurosis shortens uniformly and that the shear in the RF-VI aponeurosis showed three different patterns. Furthermore, we found considerable deformation of muscle in the axial plane during contraction, which can affect anatomic measures obtained from a single location. The methodological testing showed that the contraction-induced deformation of muscles must be considered when performing quantitative analysis in either axial or sagittal planes.

Previously, our laboratory has applied the VE-PC MRI method to examine calf muscle function during isometric contractions (7–9, 13, 16, 22). Our laboratory has reported compressive strain in the distal aponeurosis of the soleus but tensile strain in the midaponeurosis. As demonstrated theoretically, the force along the aponeurosis varies as a function of

![Fig. 6. Classification of the aponeurosis behavior into 3 categories based on relative movement between VI and RF side (A) and strain in RF (black bars) and VI (white bars) (B). DI, DII, and DIII refer to the 3 types of displacement, and SI, SII, and SIII refer to the 3 types of shear strain identified from the data (see Tissue displacement and strain in METHODS) Positive values indicate stretching and negative shortening. The individual subjects in both panes can be identified by the numbers below the bars.](http://jap.physiology.org/doi/10.1152/jappl.00351.2008)
distance from the tendon (6). In light of this theory, our laboratory’s previous results showing nonuniform strain in the soleus aponeurosis (9) can be further speculated; the data showed compressive strain in the distal aponeurosis while the theoretical model shows minimal forces in this location. The greatest tensile strain of 2.5% was found in midregion of the aponeurosis (the most proximal part was not measured) while greatest forces were modeled in the proximal region.

Contrary to the earlier observations by Finni et al. (9), the tensile strain along the aponeurosis in this study was uniform although the displacement showed heterogeneity. Our hypothesis was formed based on the previous findings of nonuniform strain in the soleus aponeurosis during isometric contractions of 20–40% MVC. Thus the different result in the QF may arise because of different muscle, type of contraction, and a smaller load used in the present study.

In the present study, where only a load corresponding to 5 kg was resisting the knee extension, the tensile strain in the aponeurosis was very low. On average the aponeurosis shortened 1.6% during the concentric contraction. This result, the aponeurosis shortening when force is applied is in conflict with the monotonic length-tension relationship reported for tendon structures. However, when we consider the facts that 1) the aponeurosis is anatomically tightly coupled with the muscle and 2) the entire muscle shortened during the concentric contraction, the result is logical. Previously, aponeurosis shortening during active contractions has been reported by Zuurbier et al. (24), Maganaris et al. (19), and Lieber et al. (17).

Interestingly, the shear between RF and VI showed three individual patterns. When the classification into three types of shear was done based on the maximum displacement or strain, the groups contained different populations (Fig. 6). In the literature RF and VI have been described to have insignificant mechanical ties while strong mechanical relationships have been suggested between VL and RF (10). The minor mechanical ties between RF and VI presumably allow relatively free movement between the muscles. Thus to explain our results of the individual relative movement between these muscles, we hypothesized that their behavior was related to the either neural activity or anatomy.

We first sought answer to the individual patterns from the relationships between the shear type and muscle EMG activity level, but it could not explain the individual behavior. The large variation in the EMG percentage values (%MVC) may arise from the different position during the MVC knee extension, which was done in sitting position with hip flexed compared with the cyclic knee extension-flexion that was performed in a prone position. In the prone position, the intramuscular coordination may have been different from sitting, but we did not control this factor in this study. On the other hand, the cross-correlation coefficients and phase shift gave no reason to think that the RF and the vasti are functioning differently between individuals.

We also examined the relationships between anatomy and the three types of shear. The best explanation found for the individual type of RF-VI aponeurosis movement was the relative volume of the RF to the vastus muscles (Fig. 7). The correlation suggests that if one has a greater volume (and CSA) of RF in relation to the vastus muscles, then the RF is likely to move more than the vasti. This may be because of the greater strength brought by the greater CSA. However, because we did not measure muscle fascicle lengths, the relationship between the architecture (namely physiological cross-sectional area) and the individual function cannot be concluded without confirmatory measurements.

Another important result of this study was that the changes in muscle thickness during contraction depend on the location of observation. The first widely noted observation on the fact that muscle thickness does not remain constant as previously assumed by many models was published by Maganaris et al. (18). Using 2-D ultrasonography, they showed that in the triceps surae muscle the thickness increased or did not change depending on the muscle. The present study takes the problem of modeling muscle function further by adding another dimension. Clearly, the muscle deforms in three dimensions where the adjacent muscles affect the deformation. For example, when examined from the lateral view (frontal plane), the VL muscle thickness did not change, but changing the site of measurement to the 45° plane between sagittal and frontal gave significant changes. Because MRI and ultrasonography are both reliable tools to measure muscle thickness (4) the consequence of three-dimensional (3-D) muscle deformation on the thickness measurement should also be considered in ultrasound measurements.

Currently, it is not known how well a single site of measurement reflects the overall behavior of the muscle during training, aging, or immobilization, for example. Our results show that both a simple measure such as muscle thickness and more complex measures of aponeurosis behavior may depend on the site of observation. Ongoing studies are currently examining the effect of strength training on muscle thickness at several sites of measurement.

Another interesting comparison of the site of observation is when the velocity and displacement of the mid muscle belly or aponeurosis (i.e., deep or superficial border of the muscle) was
examined. As expected, when measured from the mid muscle belly at distal and proximal locations, shortening of muscle occurred during concentric contraction. However, depending on the site of measurement in the proximity of the RF-VI aponeurosis, the muscle (or in fact the aponeurosis) appeared to lengthen or shorten. This nonuniformity may reflect different forces applied along the length of the aponeurosis during a low-load contraction (9).

Regarding muscle deformation in the lower leg, our laboratory previously reported that the contraction causes different changes along the length of the muscle: the CSA decreases in the distal muscle and increases in the proximal muscle (13). The point where no change occurred in CSA located at ~75% of the muscle length from distal end. Although we were not able to assess this value for QF muscle (only 2 slices were measured during contraction), it may be assumed similar to that in the lower leg. We hypothesize that this location would be least affected by longitudinal deformation of the muscle when anatomic observations are made. Furthermore, also the lateral deformations may be smaller at the proximal muscle, thus making the observations from single point more reliable. These hypotheses require further studies to be confirmed because we have not measured axial sections either from lower or upper leg during contractions.

The prone position of the subjects during the MRI scans causes pressure on the quadriceps muscle, thus potentially affecting the shape of the muscle. However, we tested the effect of position on the muscle deformation during contraction and confirmed that the muscles undergo similar deformation in both prone and supine position. Also, the force that the muscle can produce was not affected by the position, at least not at the submaximal levels of muscle contraction used in the present study. Furthermore, while there is friction between the scanner bed (and coil) and the skin, the muscles can move relative to the skin as they do in any other position (23).

**Methodological Testing**

The results showed that the pseudo-2-D tracking of movement from unidirectionally encoded velocities is possible in close proximity of identifiable structures such as an aponeurosis. The pseudo-2-D tracking involves correction for the lateral displacement using the magnitude images as a reference. This enables faster scans and/or higher temporal resolution than if the velocity was encoded in 3-D. The automatic digitizing algorithm used in the pseudo-2-D tracking test was far from optimal, but it seems to give values comparable to manual digitizing with the significant advantage of consistency. The points are always selected according to set criteria.

Comparison of the velocities in axial and sagittal planes shows that the differences in velocities are well within the range of within-session variance. Moreover, calculating the displacement from the flow through the voxels in the ROI gives displacement values reasonably close to those derived from tracking the voxels in sagittal plane throughout the cycle. This would suggest that for clinical purposes, to facilitate the analysis and to reduce the number of required scans, calculating the displacement from the through-plane flow in axial images is adequate and justified.

**Conclusions**

Contrary to our hypothesis, we found uniform tensile strain along the aponeurosis in both RF and VI during low-load contractions while the displacement demonstrated heterogeneity. The present results further showed that the large individual behavior of the QF aponeurosis may be explained by structure-function relationships. Importantly, the effect of contraction on muscle thickness and aponeurosis behavior is different when measured at different locations. These observations emphasize the fact that single-point measurements may not always reflect the overall deformation of the aponeurosis or muscle.

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