ELSEVIER

Journal of Biomechanics 40 (2007) 3261–3266

JOURNAL OF BIOMECHANICS

www.elsevier.com/locate/jbiomech
www.JBiomech.com

Short communication

Muscle geometry affects accuracy of forearm volume determination by magnetic resonance imaging (MRI)

Carolyn M. Eng[a], Geoff D. Abrams[a], Laura R. Smallwood[a], Richard L. Lieber[a], Samuel R. Ward[a,b,*]

[a]*Departments of Orthopaedic Surgery and Bioengineering, University of California, San Diego, CA, USA*
[b]*Department of Radiology, University of California and Veterans Administration Medical Centers, San Diego, CA, USA*

Accepted 8 April 2007

Abstract

Upper extremity musculoskeletal modeling is becoming increasingly sophisticated, creating a growing need for subject-specific muscle size parameters. One method for determining subject-specific muscle volume is magnetic resonance imaging (MRI). The purpose of this study was to determine the validity of MRI-derived muscle volumes in the human forearm across a variety of muscle sizes and shapes. Seventeen cadaveric forearms were scanned using a fast-spoiled gradient echo pulse sequence with high isotropic spatial resolution (1 mm^3 voxels) on a 3T MR system. Pronator teres (PT), extensor carpi radialis brevis (ECRB), extensor pollicis longus (EPL), flexor carpi ulnaris (FCU), and brachioradialis (BR) muscles were manually segmented allowing volume to be calculated. Forearms were then dissected, muscles isolated, and muscle masses obtained, which allowed computation of muscle volume. Intraclass correlation coefficients ($ICC_{2,1}$) and absolute volume differences were used to compare measurement methods. There was excellent agreement between the anatomical and MRI-derived muscle volumes (ICC = 0.97, relative error = 12.8%) when all 43 muscles were considered together. When individual muscles were considered, there was excellent agreement between measurement methods for PT (ICC = 0.97, relative error = 8.4%), ECRB (ICC = 0.93, relative error = 7.7%), and FCU (ICC = 0.91, relative error = 9.8%), and fair agreement for EPL (ICC = 0.68, relative error = 21.6%) and BR (ICC = 0.93, relative error = 17.2%). Thus, while MRI-based measurements of muscle volume produce relatively small errors in some muscles, muscles with high surface area-to-volume ratios may predispose them to segmentation error, and, therefore, the accuracy of these measurements may be unacceptable.

Keywords: Magnetic resonance imaging; Muscle volume; Muscle architecture

1. Introduction

The ability to quantify skeletal muscle dimensions accurately *in vivo* is becoming increasingly important. Muscle size (i.e., mass or volume) improves the specificity and predictive power of biomechanical models of the musculoskeletal system which often rely on this parameter to estimate mechanical force (Holzbaur et al., 2005). It may also be useful to determine the efficacy of strength training (Harridge et al., 1999), adaptation to space flight (LeBlanc et al., 2000), and response to aging (Overend et al., 1992).

One injury that is particularly devastating to patient function is spinal cord injury. Depending on the level and severity of the injury, sensory and motor innervation to a muscle may be completely lost, resulting in a non-functional muscle. However, motor loss may be partially restored by transferring the distal tendon of a healthy, functional muscle to the distal tendon of a non-functional muscle. The net effect is to "power" the non-functional tendon with a healthy muscle to restore mobility. This intervention is commonly used after spinal cord injury where some muscles remain functional, while others are impaired or non-functional (Riordan, 1983). Pronator teres (PT) is a commonly used donor muscle in these tendon

*Corresponding author. Skeletal Muscle Physiology Laboratory, Departments of Orthopaedic Surgery, Bioengineering, and Radiology, 9500 Gilman Drive (mail code 9151), La Jolla, CA 92093, USA.
Tel.: +1 858 534 4918; fax: +1 858 552 4381.
E-mail address: srward@ucsd.edu (S.R. Ward).

0021-9290/$ - see front matter © 2007 Elsevier Ltd. All rights reserved.
doi:10.1016/j.jbiomech.2007.04.005

transfer surgeries as its relatively high spinal innervation leaves it functional after C6 spinal cord injury. It can be used to restore wrist extension by transfer to extensor carpi radialis brevis (ECRB) and thumb extension to extensor pollicis longus (EPL) (Riordan, 1983). Flexor carpi ulnaris (FCU) is a candidate for surgical restoration of digital extension following high radial nerve palsy (Zachary, 1946), and brachioradialis (BR) can be used to restore thumb flexion in patients with tetrapelegia (Hentz et al., 1983). Literature regarding optimizing the outcome of tendon-transfer surgeries has focused on pre-operative decisions to match muscle function of transferred muscles (Brand et al., 1981) and intraoperative techniques to reattach muscles at their optimal length (Fridén and Lieber, 2002). However, there is little literature documenting muscle function post-operatively and the ability to serially measure muscle size would provide valuable insights into muscle function in the weeks and months that follow surgery.

Techniques currently used to measure muscle size *in vivo* are not specific to single muscles and their accuracy depends on a number of uncontrollable variables. For example, the accuracy of external anthropometric measures (Jones and Pearson, 1969) varies depending on the subject, the geometry of the limb of interest, and the amount of subcutaneous fat (Rice et al., 1990). The resolution of bioelectric impedance analysis (Brown et al., 1988) and dual-energy X-ray absorbitiometry (DXA; Shih et al., 2000) only provides an estimate of total limb muscle mass and is not single muscle specific.

A number of noninvasive techniques offer the potential for measuring a subjects' musculoskeletal dimensions. Although several imaging modalities have been utilized for this purpose (i.e., computed tomography (CT) and ultrasound (US)), magnetic resonance imaging (MRI) offers distinct advantages over these modalities. MRI images provide high contrast of muscle, fat, and connective tissue, allowing delineation of muscle borders. MRI does not expose subjects to ionizing radiation and thus may be advantageous to CT in longitudinal studies where subjects require multiple scans. Furthermore, MRI provides a large field of view (FOV) relative to US, which enables visualization of whole muscles and limbs. However, the accuracy of measuring muscle size (volume) with MRI has not been well established. Previous attempts to validate MRI-based muscle volume measurements relied on phantom calibrations (Tracy et al., 2003) or a wide range of muscle sizes (Fukunaga et al., 2001; Scott et al., 1993) which do not establish the accuracy of serial volumetric measurements under realistic conditions. The most rigorous validation (Tingart et al., 2003) suggested very accurate MR-based volume measurements (errors ~4%) in rotator cuff muscles, yet the muscles examined in this study have well-defined bony compartments, and those that did not (infraspinatus and teres minor) were combined into a single volume measurement. This oversimplified approach minimizes muscle identification errors. Additionally, high-field strength MR systems, which promise better signal-to-noise ratios (SNR) and higher spatial resolution, but may also have larger spatial distortions, which have not been studied.

To establish the accuracy of measuring muscle volumes *in vivo*, we characterized the hardware and muscle-specific errors associated with measuring muscle volumes in the forearm using a commercially available high-field strength MRI system. These experiments are unique in that they establish fixed and modifiable sources of measurement error in perhaps the most complex extremity system (wrist and hand) examined to date.

2. Methods

Forearm specimens (distal third of the humerus to the carpals) were obtained from 17 fixed cadavers (82±8 years; PT, ECRB, and EPL: $n = 10$, FCU: $n = 7$, BR: $n = 6$). Prior to imaging, it was determined that a 35 cm imaging FOV would allow the region between the distal carpal row and the proximal humeral epicondyles to be visualized in all specimens. To characterize the spatial distortions produced by magnetic field inhomogeneities within this FOV, a 48 cm long, 4.3 cm diameter

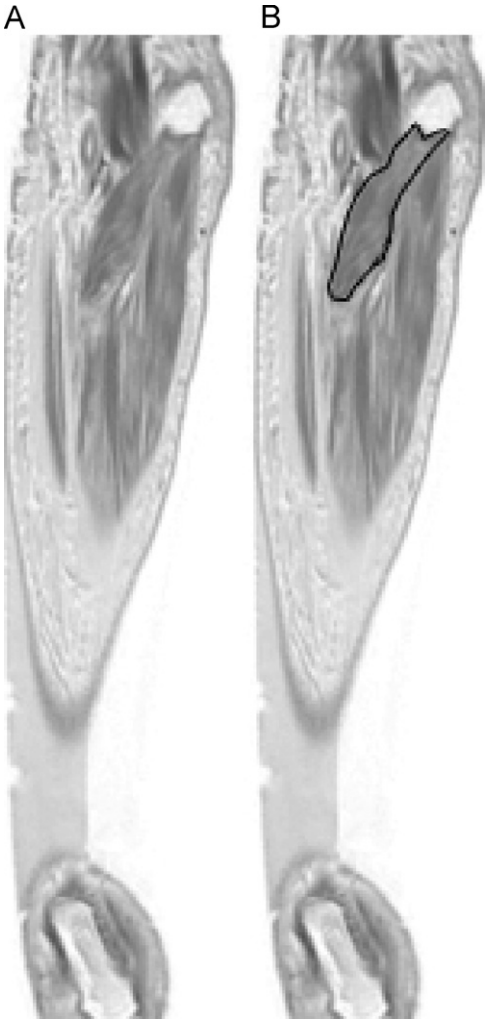

Fig. 1. Sagittal image (inverted, fat-suppressed FSPGR) of the forearm showing PT muscle without (A) and with (B) the muscle border outlined. Manual segmentation required outlining each muscle in all MR images that contained the muscle in order to reconstruct muscle volume.

water-filled plastic pipe was imaged and reformatted in the three cardinal planes. Measurements of maximum pipe width in the sagittal (acquisition) and coronal planes were made in 1 cm increments along the length of the pipe to generate a first approximation of measurement error. These pilot experiments revealed that linear measurement errors were small in the acquisition plane (4% maximum, $n = 35$ measurements) and relatively large out of the acquisition plane (25% maximum, $n = 35$ measurements) at the ends of the FOV.

Given these non-uniform measurement errors, it was apparent that quantifying hardware-specific volume errors within our chosen FOV dimension may be important. Errors were quantified by scanning 15 mL water phantoms at known positions within the 35 cm FOV, calculating their volumes from the MR images and comparing them to the known volume. The 15 mL volume was chosen because it was representative of the PT, ECRB, EPL, FCU, and BR muscle size (mean muscle volume = 19.9 ± 8.3 mL).

Prior to dissection, MR images of each forearm were obtained using a 3.0 T Signa MR imaging system (GE Medical Systems, Milwaukee, WI). Images were acquired in the sagittal plane using a three-dimensional (3D) fast-spoiled gradient recalled echo (FSPGR) pulse sequence (TR 9.2, TE 3.9, TI 24, flip angle 30°, NEX 1, FOV 35 × 35 cm, matrix 352 × 352, slice thickness 1.0 mm, and an eight-channel head coil). These parameters yielded an isotropic voxel dimension of 1 mm^3, good contrast between muscle and fascia, and an acquisition time that would be reasonable for living subjects (7.5 min).

After image acquisition, muscle volume was measured from image data sets using manual segmentation (Analyze version 7.0, Analyze Direct, Lenexa, KS; Fig. 1). The PT, ECRB, EPL, FCU, and BR muscles were manually segmented in the three cardinal planes. In addition, several volumes were re-sampled in an oblique plane to visualize the muscles of interest more clearly. Toggling between planes allowed the operator to construct a single volumetric mask of a muscle from multiple views. This ensured 3D accuracy of muscle segmentation. If there was a discrepancy in the defined borders of the muscle between orientation planes, the segmentation was reconciled manually between planes until the borders of the muscles agreed in all three orientation planes.

To establish interexaminer reliability, a second investigator segmented and measured the volume of each muscle. Volumes were compared between investigators using the intraclass correlation coefficient ($ICC_{2,1}$) and percent difference. These data confirmed that there was excellent agreement between examiners ($ICC_{2,1} = 0.97$, average percent difference 8.8%).

Once each muscle was segmented, volume and surface area were calculated. Surface area was determined in the Analyze software by calculating the number of pixels in both end slices of an object and then adding the number of the pixels enclosing the perimeter of each slice of the muscle and multiplying by the pixel dimension (mm^2). Volume was calculated by adding the number of voxels contained in the muscle and multiplying by the voxel dimension (1 mm^3).

After scanning and image processing, forearms were skinned and the PT, ECRB, EPL, FCU, and BR muscles were removed. Muscles were dissected of excess fat and fascia and external tendons were detached. Each muscle was weighed (to the nearest 0.01 g) and its volume was calculated using a density value appropriate for the method of fixation (Ward and Lieber, 2005).

The degree of absolute agreement between dissection- and MR-based volume measurements was determined using the ICC. Relative error between measurement techniques was calculated using percent difference. P values < 0.05 were considered significant and all values are reported as mean ± S.E. unless otherwise noted.

3. Results and discussion

Phantom testing determined that a 35 cm FOV yields volume errors as high as 21% at the ends of the FOV (Fig. 2). However, forearm muscles may not all be subjected to this same degree of error because, unlike the phantom vials, muscles spanned a range of horizontal distances within the FOV. Therefore, although these data were useful to determine hardware-associated errors, they likely overestimated error that would be observed *in vivo*.

There was good agreement between MRI and dissection-based volumes when comparing all muscles ($ICC_{2,1} = 0.97$ and percent difference = 12.8%; Fig. 3A). When considering

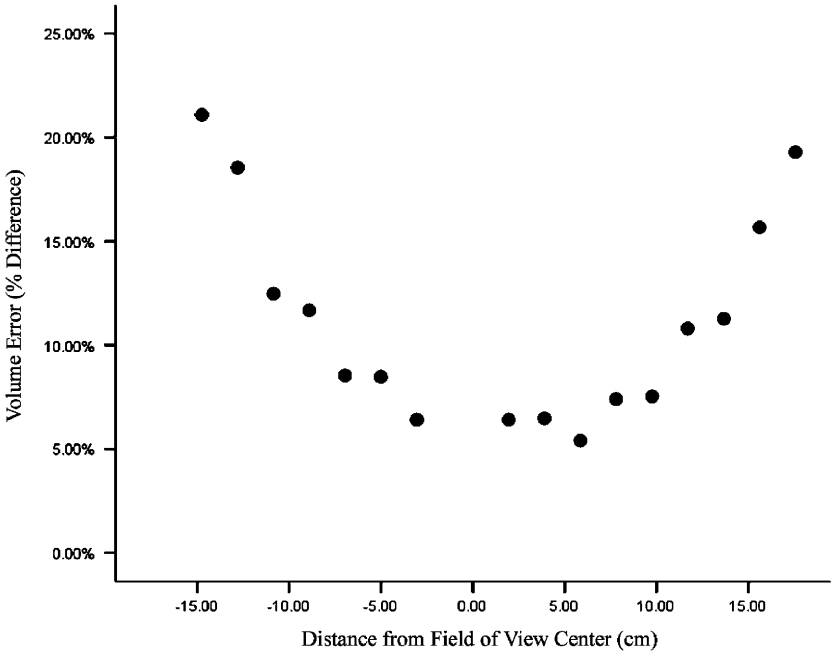

Fig. 2. Scatter plot of volumetric error as a function of the distance from the center of the scanning field of view (35 cm). Error was determined by comparing MR-based volume to the known phantom volume (15 mL).

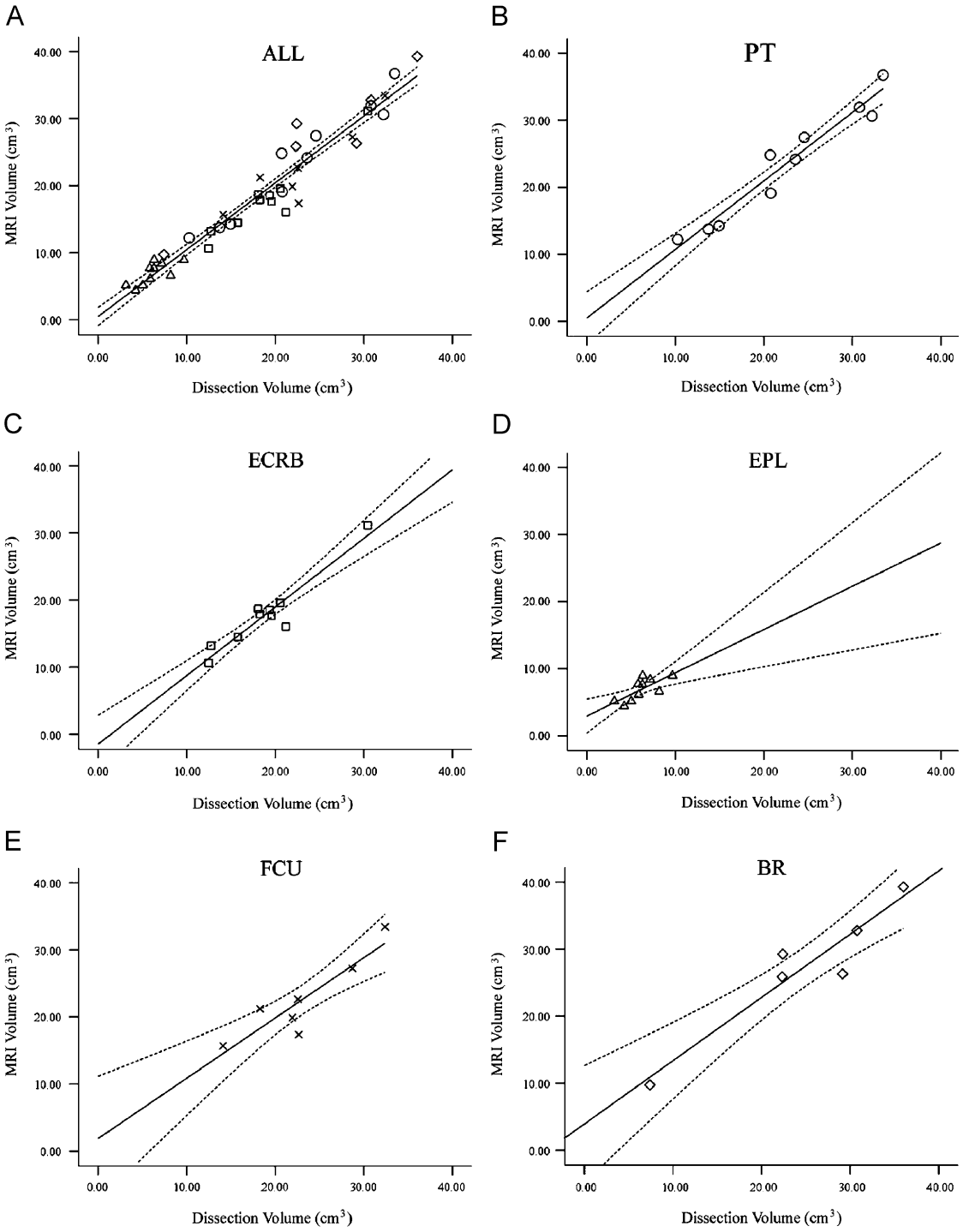

Fig. 3. Scatter plot of muscle volume calculated from MRI segmentation (MRI Volume) versus muscle volume determined from direct dissection of muscle (Dissection Volume) of (A) all muscles, (B) pronator teres muscle (PT; circles), (C) extensor carpi radialis brevis (ECRB; squares), (D) extensor pollicis longus (EPL; triangles), (E) flexor carpi ulnaris (FCU; crosses), and (F) brachioradialis (BR; diamonds). Solid lines represent the regression equations and the dashed lines represent the 95% confidence interval of the regression equation. All graphs are shown across the same range of independent variables for clarity. In practice, regression relationships would only be used across the range of independent variables appropriate to that muscle.

individual muscles, there was excellent agreement between measurement techniques for several muscles of the forearm: PT ($ICC_{2,1} = 0.97$ and percent difference = 8.4%; Fig. 3B), ECRB ($ICC_{2,1} = 0.93$ and percent difference = 7.7%; Fig. 3C), and FCU ($ICC_{2,1} = 0.91$ and percent difference = 9.8%; Fig. 3E). Contrary to these muscles, there was only fair agreement between measurement techniques for EPL ($ICC_{2,1} = 0.68$ and percent difference = 21.6%; Fig. 3D) and BR ($ICC_{2,1} = 0.93$ and percent difference = 17.2%; Fig. 3F).

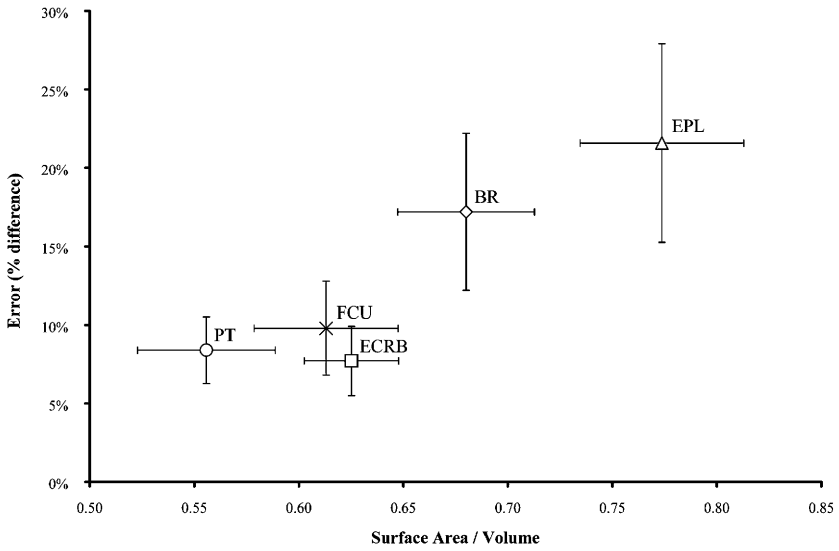

Fig. 4. Scatter plot of error difference between MRI and dissection-based measurements versus surface area:volume ratio for pronator teres (PT; circle), extensor carpi radialis brevis (ECRB; square), extensor pollicis longus (EPL; triangle), flexor carpi ulnaris (FCU; cross), and brachioradialis (BR; diamond). Data shown as mean ± S.E.M.

Across all muscles, MRI volume measurements were accurate. However, the accuracy of this technique was muscle specific as the percent difference between MRI and dissection measurement techniques ranged from 7.7% to 21.6%. One possible explanation for this variability among muscles is that the resolution of the images is not high enough to resolve the borders of smaller muscles. However, absolute error between MRI- and dissection-based volume corresponded with total muscle volumes. For example, the EPL and BR had the smallest and largest volume and absolute volume error, respectively. Furthermore, the resolution of the images was $1\,mm^3$, or about 1/6170th of the total volume of EPL. Therefore, it is unlikely that the resolution of the MR images led to the error variability measured among muscles.

Although it is intuitively appealing to suggest that these error differences can be attributed to spatial distortions as observed in the phantom testing, our results directly contradict this assertion. The EPL muscle is the most centrally located muscle within the forearm, and therefore would be subjected to the lowest distortion in the FOV (Fig. 2), yet this muscle yielded the greatest error. PT, BR, and ECRB have a positional bias toward the proximal forearm, and therefore, would be subjected to the highest spatial distortions. Another possibility regarding muscle-specific accuracy is that segmentation errors are responsible for these differences. Manual segmentation requires the border of each muscle to be visually defined in each image. Therefore, a long muscle with a small diameter would require relatively more decisions to be made defining the border compared to a short muscle with a larger diameter. To quantify the shape characteristics of each muscle, the surface area-to-volume ratio of each muscle was calculated. When muscles were compared, there was a strong relationship between high surface area-to-volume ratios and measurement error ($R^2 = 0.85$, $P < 0.05$). The EPL and BR muscles had larger surface area-to-volume ratios and larger measurement errors compared to PT, ECRB, and FCU (Fig. 4). These data suggest that muscle shape and manual segmentation errors are likely to be responsible for the unacceptably high volume errors in these muscles and reinforce the need to validate volumetric measurements on a muscle-by-muscle basis.

4. Summary

These data provide the first quantitative evidence that high-resolution MRI can accurately quantify forearm muscle volume. The volume errors observed in this study were the result of both spatial distortions incurred by acquiring images with a relatively large FOV and manual segmentation errors. In the latter case, muscle shape likely influenced the magnitude of volumetric errors. Modifiable measurement errors, such as FOV size can be minimized simply by matching the FOV dimension to the region of interest, or placing the muscle of interest in the center of the FOV, where field distortion is minimized.

Although not directly addressed by this study, we would like to emphasize that knowledge of 3D gross anatomy is required for accurate muscle segmentation. This is particularly important in the forearm as muscles are closely packed with thin fascial boundaries and often run obliquely to the imaging planes. We have demonstrated that there is excellent agreement between examiners of similar anatomical expertise here, but care should be used when extrapolating our data to other individuals or muscles.

MRI is a useful tool for the measurement of muscle volume *in vivo*. However, given the errors observed in this study (~10%), longitudinal studies attempting to measure

effect sizes of less than 10% should carefully consider possible limitations of low statistical power. Additionally, although muscle volume is a key determinant of physiological cross-sectional area, the functional relevance of this parameter can be questioned particularly when muscle fiber length may change. Future studies are necessary to validate methods of obtaining other muscle morphometric parameters including fiber length and pennation angle, which will allow the direct computation of muscle architectural properties.

Conflict of interest

The authors confirm that the publication of this paper involves no conflict of interest of any kind.

Acknowledgments

The authors gratefully acknowledge the Body Donation Program and the Anatomical Services Department at the University of California, San Diego. Support for this study was provided by the National Institutes of Health (Grants HD048501 and HD050837).

References

Brand, P.W., Beach, R.B., Thompson, D.E., 1981. Relative tension and potential excursion of muscles in the forearm and hand. Journal of Hand Surgery (American) 3A, 209–219.

Brown, B.H., Karatzas, T., Nakielny, R., Clarke, R.G., 1988. Determination of upper arm muscle and fat areas using electrical impedance measurements. Clinical Physics and Physiological Measurement 9, 47–55.

Fridén, J., Lieber, R.L., 2002. Mechanical considerations in the design of surgical reconstructive procedures. Journal of Biomechanics 35, 1039–1045.

Fukunaga, T., Miyatani, M., Tachi, M., Kouzaki, M., Kawakami, Y., Kanehisa, H., 2001. Muscle volume is a major determinant of joint torque in humans. Acta Physiologica Scandinavica 172, 249–255.

Harridge, S.D., Kryger, A., Stensgaard, A., 1999. Knee extensor strength, activation, and size in very elderly people following strength training. Muscle and Nerve 22, 831–839.

Hentz, V.R., Brown, M., Keoshian, L.A., 1983. Upper limb reconstruction in quadriplegia. Functional assessment and proposed treatment modification. Journal of Hand Surgery (American) 8A, 119–131.

Holzbaur, K.R., Murray, W.M., Delp, S.L., 2005. A model of the upper extremity for simulating musculoskeletal surgery and analyzing neuromuscular control. Annals of Biomedical Engineering 33, 829–840.

Jones, P.R., Pearson, J., 1969. Anthropometric determination of leg fat and muscle plus bone volumes in young male and female adults. Journal of Physiology 204, 63–66.

LeBlanc, A., Lin, C., Shackelford, L., Sinitsyn, V., Evans, H., Belichenko, O., Schenkman, B., Kozlovskaya, I., Oganov, V., Bakulin, A., Hedrick, T., Feeback, D., 2000. Muscle volume, MRI relaxation times (T2), and body composition after spaceflight. Journal of Applied Physiology 89, 2158–2164.

Overend, T.J., Cunningham, D.A., Paterson, D.H., Lefcoe, M.S., 1992. Thigh composition in young and elderly men determined by computed tomography. Clinical Physiology 12, 629–640.

Rice, C.L., Cunningham, D.A., Paterson, D.H., Lefcoe, M.S., 1990. A comparison of anthropometry with computed tomography in limbs of young and aged men. Journal of Gerontology 4, 175–179.

Riordan, D.C., 1983. Tendon transfers in hand surgery. Journal of Hand Surgery (American) 8, 748–753.

Scott, S.H., Engstrom, C.M., Loeb, G.E., 1993. Morphometry of human thigh muscles. Determination of fascicle architecture by magnetic resonance imaging. Journal of Anatomy 182, 249–257.

Shih, R., Wang, Z., Heo, M., Wang, W., Heymsfield, S.B., 2000. Lower limb skeletal muscle mass: development of dual-energy X-ray absorptiometry prediction model. Journal of Applied Physiology 89, 1380–1386.

Tingart, M.J., Apreleva, M., Lehtinen, J.T., Capell, B., Palmer, W.E., Warner, J.J., 2003. Magnetic resonance imaging in quantitative analysis of rotator cuff muscle volume. Clinical Orthopaedics and Related Research Oct, 104–110.

Tracy, B.L., Ivey, F.M., Jeffrey Metter, E., Fleg, J.L., Siegel, E.L., Hurley, B.F., 2003. A more efficient magnetic resonance imaging-based strategy for measuring quadriceps muscle volume. Medicine and Science in Sports and Exercise 35, 425–433.

Ward, S.R., Lieber, R.L., 2005. Density and hydration of fresh and fixed skeletal muscle. Journal of Biomechanics 38, 2317–2320.

Zachary, R.B., 1946. Tendon transplantation for radial paralysis. British Journal of Surgery 33, 358–364.