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RECTUS FEMORIS MUSCLE VOLUME ESTIMATE BY ULTRASONOGRAPHY

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SUMMARY
Imaging techniques allow in vivo measurement of parameters studied in vitro or ex vivo and thus greater reliability regarding muscle mechanics in real situations. Some muscle architecture parameters are directly related to force production, mainly the volume. It can be estimated by the sum of segmental volumes (product of the sum of cross-sectional areas and the distance between them) of sequential images obtained from magnetic resonance imaging. Ultrasound has also been used to obtain the volume utilizing regression equations or the same method as in magnetic resonance images. However, the prolonged time for acquisition and processing many images can be a limiting factor. Therefore, this paper presents a model based on the Cavalieri´s principle to estimate the rectus femoris volume. Different estimations of a phantom volume were performed with 2, 3, 4, 5, 6, 7 and 8 cross-sectional images to define a protocol with the minimum number of images sufficient for a reliable estimate. The 4 slices prediction approached the experimental value with a mean error of 1.12%.

INTRODUCTION
Imaging techniques such as magnetic resonance imaging (MRI), computed tomography (CT) and ultrasonography (US) are used in the skeletal muscle characterization to provide more accurate information for the diagnosis and treatment of injuries, as well as monitoring different types of training, rehabilitation programs and the effects of aging. These techniques allow in vivo measurements of variables commonly studied in vitro or ex vivo, increasing the reliability regarding the muscle mechanics [1,2].

Some muscle architecture parameters, especially the volume, are directly related to its force production capacity and can be used to estimate the individual contribution of a muscle in a joint torque [2]. Muscle volume (MV) can be estimated by the sum of segmental volumes (sum of cross-sectional areas multiplied by the distance between them) obtained from MRI [3].

Muscle thickness and anatomical cross-sectional area (CSA), which can be measured in US images, associated with other anthropometric variables are used in regression equations to estimate MV. These equations were validated through in vivo MRI images or in vitro hydrostatic weighing for different muscle groups, such as elbow flexors and quadriceps [3,4]. However, it is inappropriate to extrapolate these equations to distinct populations the ones of the studies.

Volume can also be estimated with US as in MRI, and has some advantages such as lower cost, easy handling and portability [1]. However, the olonged time on the acquisition of a series of images can be a limiting factor as the operator must manually perform several small accurate probe displacements. Thus, as in the MRI, the operational cost of measuring the CSAs of many images is high.

Nevertheless, the shape of an irregular solid can be replaced by a symmetric equivalent with same height and CSAs whose volume equation is known. This is an acceptable approximation of the Cavalieri's principle with high practical application. Thus, this work shows an equivalent model to predict rectus femoris volume, since it is a superficial muscle with great functional importance and easily distinguishable in US images. The analysis of model behavior as a function of the slices (images) quantity in a phantom was made to define a protocol with the optimum number of images for a reliable estimate.

METHODS
The rectus femoris phantom production followed the Maggi`s protocol [5] modified by the addition of graphite, aiming to adjust the granulation and to generate images similar to the muscle in vivo. The phantom was produced with size and shape based on anthropometrical characteristics of a male volunteer (Figure 1).

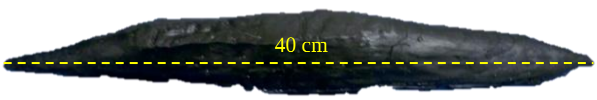

Figure 1: Rectus femoris phantom.

Characterization of two acoustic parameters (propagation speed and attenuation coefficient) was performed to verify phantom similarities to skeletal muscle, given their influence on image formation and measurements [5].

The ultrasound probe (SonixMDP, Ultrasonix, British Columbia, Canada), operating in 14 MHz, was immersed in a water tank to avoid phantom compression and increase acoustic coupling. The phantom was fixed and the transducer was manually displaced in 1 cm steps, generating 38 cross-sectional images.

The images were sent to a microcomputer for further analysis. An experienced evaluator performed three independent CSA measurements in each image of the phantom with the software ImageJ (National Institutes of Health, Maryland, USA). The average was used as the representative value of each area image.

The phantom was immersed in a graduated test-tube (20 ml resolution) and its volume measured by the displacement of the water column. This procedure was performed ten times, and the mean represented the measured volume. The volume estimated by US was calculated by the sum of segmental volumes of truncated cones given by Equation 1. Each truncated cone was generated from pairs of consecutive circles equivalent to the areas measured in the images.

$$Volume = \frac{\pi h}{3}\sum_i (r_i^2 + 2r_i r_{i+1} + r_{i+1}^2) \qquad (1)$$

where h is the distance between two consecutive images and r_i and r_{i+1} the equivalent radius.

Different volume estimations were performed with 2, 3, 4, 5, 6, 7 and 8 slices guided by results of a previous study [6] which showed an overestimate for larger quantities. For instance, to combine 4 images, the phantom was first divided into 4 equally spaces regions. The 38 images were distributed in respectively 10, 9, 9 and 10. Then all possible combinations of a US image of each region were made. The volume was estimated for each of such combinations and its average was obtained and also the mean percentage error.

RESULTS AND DISCUSSION

The acoustic properties of the phantom are in the soft tissues range (propagating speed 1492.95 $m.s^{-1}$ and attenuation coefficient 1.54 $dB.cm^{-1}.MHz^{-1}$) [5].

The phantom volume measured in the test tube was 340 ± 2.4 ml (mean ± sd). Erskine et al. [7] estimated MV in a sample with anthropometric characteristics similar to the male volunteer who served as model for the phantom used in this study. Their results (339 ± 61 ml) were close to ours.

The model underestimated the volume when 2 and 3 slices were used and overestimated with 5, 6, 7 and 8, so it is proposed a multiplier factor (Table 1). The estimate with 4 slices approached the experimental value presenting a mean error of 1.12%. It would be expected lower errors with the inclusion of more slices, since more information of the areas of the phantom would be available. Besides the smallest error, the use of 4 slices represents a reduction in the operating cost of acquiring and processing multiple images.

However, combinations with better results should still be searched as also regions of the muscle for acquisition of the best slices in order to develop a simplified protocol.

Some studies used multiple regression equations to estimate the volume of the quadriceps. Miyatani et al. [3] investigated the accuracy and reliability of volume obtained from measurements of knee extensors muscle thickness from ultrasound images. The equations presented a coefficient of determination of 0.787. However, to obtain the rectus femoris volume, one more estimate was needed (the percentage of this compartment), which could result in associated errors, in addition to the 11.1% standard error of estimate reported by the authors.

Few studies deal with the segments between successive images as truncated cones, instead of a cylinder. It would create a model closer to the real muscle shape, since, for the present study, the areas measured in all pairs of consecutive slices were different from each other (ie distinct from a cylinder). Infantolino et al. [4] reported average error of only 0.4 ± 6.9% for the vastus lateralis estimated volume (by truncated cones) of cadavers in comparison to hydrostatic weighing.

CONCLUSIONS

This study proposes a method of estimating rectus femoris muscle volume with a minimum number of ultrasonic images. The protocols with more slices presented greater errors and demanded more measurement time compared to 4 slices. Besides the accuracy, the precision must be tested in other phantoms based on individuals with different characteristics (anthropometric, age, gender, etc.).

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Table 1: Estimated volume, percentage error, multiplier factor and number of combinations in function of the images quantity.

Slices	Estimated Volume (ml)* (mean ± sd)	Percentage Error (%) (mean ± sd)	Multiplier Factor	Combinations
2	175.43 ± 49.58	48.41 ± 1.35	2.22	361
3	292.64 ± 35.69	13.93 ± 0.16	1.18	2028
4	343.81 ± 25.83	-1.12 ± 0.08	0.99	8100
5	375.18 ± 20.39	-10.35 ± 0.05	0.91	25088
6	389.13 ± 16.77	-14.45 ± 0.04	0.87	63504
7	401.69 ± 15.68	-18.15 ± 0.03	0.85	135000
8	406.15 ± 11.35	-19.45 ± 0.03	0.84	250000

* The phantom volume measured in the test tube was 340 ± 2.4 ml (mean ± standard deviation).