A METHOD OF MEASURING THE MODULUS OF ELASTICITY OF HUMAN MUSCLE TISSUE

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The viscoelastic properties of human soft tissues are measured and evaluated by means of many different coefficients and parameters introduced by the authors of published work depending on the methods and devices they have used [1-3], and it is therefore practically impossible to compare results obtained by different workers.

The object of this investigation was to develop a method of measuring mechanical parameters of human soft tissues *in vivo* which would allow the results to be obtained in generally accepted units of measurement. The biceps brachii muscle was chosen as the test object. The state of its viscoelastic properties was estimated as the modulus of elasticity E, which was calculated by the angle of slope of the deformation-load characteristic curves of the test object, which reflect the character of dependence of its deformation on the force applied to it.

Any state of elastic media can be defined (assigned) if the values of three principal parameters are given: the modulus of elasticity E, the Poisson coefficient ν, and the density ρ. However, the solution to this problem in the case of a living functioning medium is complicated by the fact that none of these parameters can be measured directly. It was therefore necessary to develop a complex method whereby the desired parameters could be obtained from the results of indirect measurement. For this purpose, as a first step we used the solution to the problem in [5] of deformation of an elastic half-space under the influence of a static external force applied to its surface through a circular die. It gives a simple equation for the value of the deformation of the object δ as a function of the force F applied to it:

$$\delta = 8 \cdot \frac{1 - \nu^2}{E \cdot a} \cdot F, \tag{1}$$

where ν is the dimensionless Poisson coefficient, a the radius of the die, and E the required modulus of elasticity. The graph of this function $\delta(F)$ in the case of viscoelastic media is always shaped like a loop, and the higher the elastic properties of the medium the steeper the slope of the loop, whereas its "aperture," i.e., the deviation of the reverse course of the graph from a straight line, reflects the degree of its viscosity.

Equation (1) can be used to calculate the modulus of elasticity of a real biological object only if a number of demands both on the object itself and on the system creating and recording its deformation, arising in response to a force applied to it, are satisfied.

1. Equation (1) is valid for a half-space and, consequently, the dimensions of the region of creation and recording of the deformation, and also the magnitude of the deformation itself (up to δ_{max}) ought to be significantly (5-10 times) less than the dimensions of the test medium (both in depth and in surface area). The same demand must naturally be satisfied also by the radius of the die.

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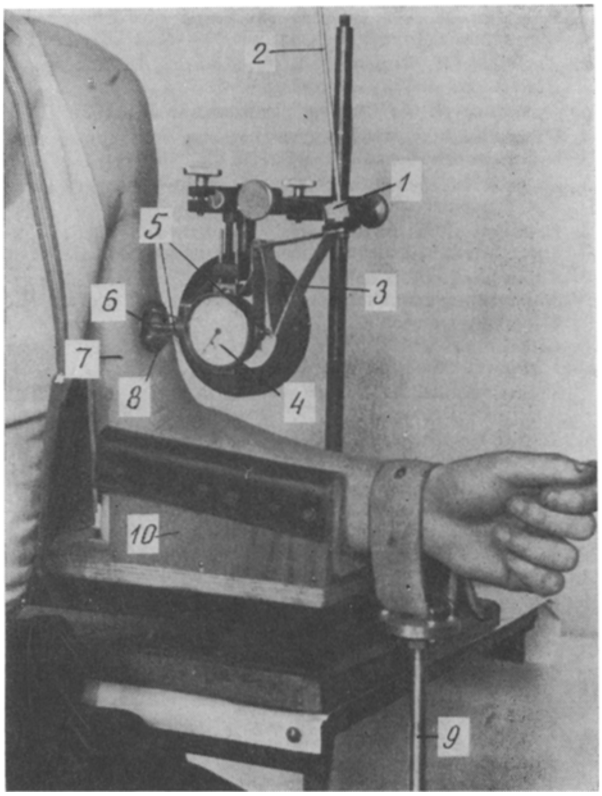

Fig. 1. General view of apparatus for creating and recording deformation-load cycle. 1) Weights; 2) guide rod; 3) rotating lever; 4) indicator head; 5) shaft; 6) interchangeable die; 7) test object; 8) supporting ring; 9) stand with set of weights; 10) armrest.

2. The function $\delta(F)$, represented by equation (1), was obtained for the case of a static task. Consequently, the magnitude of deformation must be recorded a certain time after establishment of the next value of F, and during that time all transition processes in the object must be completed and a quasisteady state established. However, since the object lives and functions, its parameters must also change. Therefore, on the one hand, the time of recording the deformation-load relationships $\delta(F)$ must be long enough, but on the other hand, it must be as short as possible so that fatigue phenomena are not able to develop in the object, for otherwise the reproducibility of the experimental results would be disturbed. In practice this time must be determined experimentally for each special problem.

3. Equation (1) applies to a linearly elastic medium. Consequently, the working region of the $\delta(F)$ curves can be taken to be their linear part, the slope of which determines the value of the required modulus of elasticity.

4. Equation (1) was obtained on the assumption that the medium is isotropic. This condition, in the case of a biological object, is practically unattainable. However, having agreed to study integral (i.e., averaged throughout the volume of the object) characteristics of its medium, we do in fact satisfy this demand for practical purposes also if we establish the dimensions of each region of the set of layers of biological tissue with which we inevitably have to deal when studying objects *in vivo*.

Since the test object (the biceps brachii muscle) has at least three "layers" (the skin with the subcutaneous fatty layer, muscle tissue itself, and bone), the half-thickness of a skin fold was first measured in the region of the biceps, and its value was subsequently plotted on $\delta(F)$ graphs, and this part of the graph was excluded from calculations of the modulus of elasticity of the muscles.

The maximal attainable deformation of the object and, consequently, the force applied to it, were naturally assigned by the thickness of the test layer, in this case muscle, and the demands of item 1. To calculate this value, the thickness of the subject's wrist was first measured (to obtain a rough estimate of the thickness of the humerus) as well as the circumference of his arm in the region of the biceps. With a combination of these two values and the thickness of the skin fold, the thickness of the layer of muscle tissue for each subject can be estimated quite accurately. The equation connecting these values has the following form:

$$d \approx \left(\frac{C}{3{,}14} - L - h\right) \text{мм}, \tag{2}$$

where d/2 is the thickness of the layer of muscle tissue, C the circumference of the biceps, h the thickness of the skin fold, and L the thickness of the wrist. The value of δ_{max}, however, according to the demands of item 1, should be $\delta_{max} \leqslant 1/5 \cdot d/2$.

Only after all the above demands have been satisfied can it be asserted that the value of E found in equation (1) and that calculated on the basis of experimentally obtained values for δ and F will in fact characterize the averaged modulus of elasticity of the given biological object, and will its value become a parameter independent of the experimenter and the system used for measurement.

The apparatus for creating and recording deformation-load cycles consisted of the following components: loading, indicator, deformation, and measured physical exertion (Fig. 1). The loading component consisted of a set of 50-g and 100-g weights, made in the form of cylinders 1, threaded on the guide rod 2, and a rotating lever 3, changing the direction of action of the force F. The indicator component consisted of a measuring head 4 taken from the standard "clock-type measuring head" instrument with simplified system for transmitting the forward movement of the shaft 5, passing through it, into rotary movement of its indicator pointer. The simple reconstruction enabled the threshold of sensitivity of the whole system to be substantially reduced, as a result of which during horizontal movement of the shaft its value was only 0.04 N for the whole 12 mm of its working path. The deformation unit consisted of the shaft 5, one end of which made mechanical contact with the arm of the rotating lever 3, whereas the other end, ending in the interchangeable die 6, made contact with the biological object 7 itself, and a supporting platform 8, shaped like a flat ring, the radius of which, to conform with the demands of item 1, was substantially greater (5-10 times) than the radius of the die. In the initial position the plane of the die lies in the plane of the supporting ring which, in turn, is adjusted to make contact with the surface of the test object. The measured exertion unit, consisting of a metal pillar 9, had a supporting platform at one end for the weights (each of 1 kg), and at the other end a device made of soft material consisting of a loop for suspending it from the subject's wrist. This component was used to create a state of tension of a certain degree in the muscles.

In the course of work the patient's upper limb, adducted to the body and flexed at a right angle at the elbow, was placed on the fixing armrest 10.

The apparatus works as follows. The supporting platform is brought up to the surface of the test object and after the routine load has been placed on the pillar, the indicator head is set at "zero." Deformation of the object is achieved by hanging the cylindrical weights (loading of the first degree was the action of the force due to the rotating lever — guide rod system, the total weight of which was 50 g) on the guide rod and recording readings of the indicator head at intervals of 2-3 sec. When δ_{max} is reached the process of unloading begins, taking readings of the instrument at the same time intervals (the value of δ_{max}, namely 12 mm, was achieved only with the muscles in the relaxed state at F = 7.4 N, and the working range of deformations for all other states of the muscles was assigned by the range of values of F from 0 to 7.4 N). After removal of the last weight (the rotating lever and guide rod) the value of the residual deformation δ_{res} is noted. The cycle lasted usually 50-80 sec and was repeated three times at an interval of 10-15 min with the same measured physical load P. On changing this load the interval was increased to 35-40 min which, according to our data, was necessary for abolishing the phenomena of fatigue.

The results were recorded in tables, after which graphs of the deformation-load characteristics were plotted, and in every case they were in the form of a loop (Fig. 2). Subsequent processing was as follows: The region of skin with a subcutaneous fatty layer was

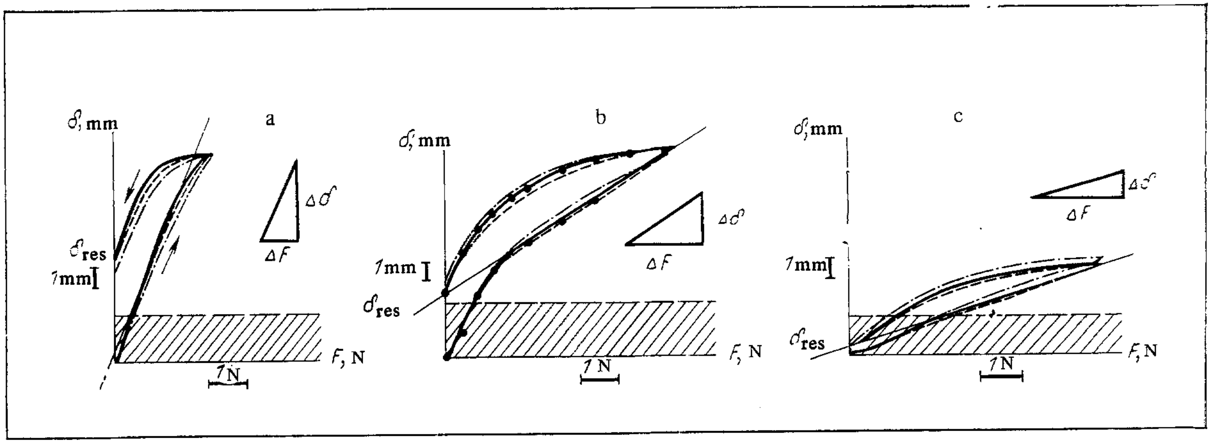

Fig. 2. Typical deformation-load curves obtained for the biceps branchii muscle. a) In relaxed state, b, c) while supporting loads of 9.8 and 39.2 N respectively. $E_a = 0.35 \times 10^5$ Pa, $E_b = 1.2 \times 10^5$ Pa, $E_c = 3.3 \times 10^5$ Pa. Abscissa, force applied to surface of muscle (in N); ordinate, magnitude of deformation of muscle beneath die of instrument (in mm).

noted, the linear part of the δ(F) graph was picked out in the region of increase of the force applied to the object, and the value of the residual deformation (δ_{res}) was recorded. The part of the graph in the region of the skin fold was excluded from subsequent calculations (in this particular task). The slope of the linear region, expressed as the ratio ΔF to Δδ, was substituted in equation (1), transformed for convenience of calculation as follows:

$$E = 1.838 \cdot 10^3 \cdot \frac{\Delta F}{\Delta \delta}, \tag{3}$$

where E is the required modulus of elasticity, expressed in pascals (or in N/m^2), ΔF the change in force in the working (linear) region of the graph, Δδ the corresponding change in deformation of the test object, and 1.838×10^3 is a dimensional coefficient, enabling values of force in units of weight of the loads (grams) and deformation in millimeters to be substituted in the equation. The radius of the die $a = 2$ mm. The Poisson coefficient ν was taken to be 0.5 on the basis of data in [4] on human skeletal muscles.

Tests were carried out on muscles in the relaxed state and in a state of isometric contraction, created in them by holding the loads P suspended from the subject's wrist. The value of P was changed with a step of 1 kg, and P_{max} was determined by the subject's capability. This approach, in our opinion, yielded a new informative parameter: dependence of the modulus of elasticity on the static load E(P), a typical graph of which is shown in Fig. 3. This is a graph of dependence of residual deformation, reflecting the degree of viscosity of the test medium on the load, which correlates completely with E(P). It is an interesting fact that the maximum of elasticity in the muscle is established simultaneously with the minimum of viscosity. The value of the load P under these circumstances is 0.25-0.3 of P_{max}. With a further increase in P both parameters keep practically the same value, with a tendency toward a decrease in elasticity and an increase in viscosity, which is evidently the first sign of fatigue. This fact was manifested most clearly in a group of subjects with poor physical development, in whom the "span" of the loop of the deformation-load curves rose sharply with loads of P = 5-7 kg, the slope of their linear region was reduced 1.5-2 times, and the value of δ_{res} increased just as sharply.

This method was used to study a group of clinically healthy men and women aged from 22 to 40 years (15 persons). The graphs in Fig. 2 show typical relationships for δ(F) obtained for three states of the biceps brachii muscle. Having three curves on each graph enables the reproducibility of the results obtained in each state to be judged. A more detailed account of the data and their analysis will be given in the next paper.

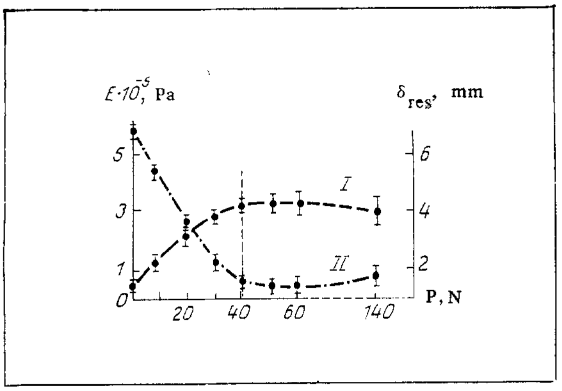

Fig. 3. Dynamics of change in modulus of elasticity E (I) and muscle strength (II) depending on degree of its isometric contraction created by supporting a load suspended from the subject's wrist. Abscissa, load (in N); ordinate: left — modulus of elasticity (in Pa), right — analog of viscosity, residual deformation (in mm).

To study the reproducibility of the parameters measured over a long period of time they were recorded, especially the modulus of elasticity and residual deformation, in a series of subjects for 1 month at intervals of 4-5 days during the morning. The results were: $E = (0.37 \pm 0.15) \times 10^5$ Pa, $\delta_{res} = 0.62 \pm 0.12$ mm, and in our opinion these demonstrate the sufficiently high reproducibility for parameters of the test objects measured *in vivo*.

The suggested method can provide a quantitative assessment of the mechanical properties of human skeletal muscles and can express these data as a generally accepted parameter, the modulus of elasticity. This method of analysis of the viscoelastic properties of human soft tissues *in vivo*, despite the structural complexity of the test object, can yield further information on their state, degree of development, and load-carrying capacity, and can be used in both clinical and sport medicine.

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