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ABSTRACT

The force-velocity and series elastic characteristics of vascular strips from the media of hog carotid arteries were determined using the method of quick release to a known afterload. Shortening velocity was a function of load, and the data could be fitted by a rectangular hyperbola using the Hill equation, \((P + a) V = b (P_o - P)\), where \(P\) is the load, \(P_o\) is the maximum isometric force developed at the optimum muscle length \(L_o\), \(V\) is velocity, and \(a\) and \(b\) are constants. The following dynamic constants were obtained for the electrically stimulated strips: estimated maximum shortening velocity for the unloaded muscle \(V_{\text{max}} = 0.12 \, \text{L}_o/\text{sec}\), \(b = 0.02 \, \text{L}_o/\text{sec}\), \(P_o = 1.80 \, \text{kg-wt/cm}^2 = 1.76 \times 10^6 \, \text{dynes/cm}^2\), and \(a/P_o = 0.18\). Maximum power output occurred at a load of 0.28 \(P_o\). The stress-strain curve for the series elastic component could be approximated by the simple exponential equation, stress = 0.034e^{0.58 \times \text{strain}}. The extension of the series elastic component at \(P_o\) was 7.2% of \(L_o\).

KEY WORDS

vascular smooth muscle mechanical properties

stress-strain relationships for the PEC and the CE of vascular smooth muscle have been described previously (1). This report describes measurements of the stress-strain curve for the SEC and the force-velocity relationship of the CE in the hog carotid media.

Methods

The preparation has been previously described in detail (1). Carotid arteries obtained from freshly slaughtered hogs\(^1\) were slit longitudinally, and 8–12 mm long strips with cross-sectional areas of 0.4–0.8 mm\(^2\) were teased from the media. The smooth muscle cells were longitudinally oriented in the strips (1) and constituted about 60% of the total strip cross section. Strips were mounted vertically between two clips. The lower clip was attached to a glass rod mounted on a micrometer for precise length adjustment. The upper clip was connected by a fine gold jewelry chain to one end of a light lever. The strips were stretched 100% of their slack length and allowed to equilibrate at this length for at least 2 hours at 37 ± 0.5°C in a tissue bath before the experiments were begun. The millimolar composition of the normal physiological salt solution was: NaCl 119, KCl 4.7, KH\(_2\)PO\(_4\) 1.18, MgCl\(_2\) 1.17, CaCl\(_2\) 1.6, NaHCO\(_3\) 14.9, dextrose 5.5, and CaNa\(_2\)ethylenediaminetetraacetate (EDTA) 0.026. The solution was bubbled continuously with 95% O\(_2\)-5% CO\(_2\) (pH 7.2–7.4). High-potassium solutions were obtained by substituting KCl for NaCl on a molar basis. The muscle

\(^1\)George H. Myers Sons, Inc., Richmond, Virginia, donated the arteries.
strips were stimulated electrically (60 Hz, 20 v/cm, for 15 seconds) at 15-minute intervals until a constant isometric tension and a constant contraction time were obtained (2-3 hours). Active tension development at several different lengths was then measured to determine the length ($L_o$) at which maximal tension ($P_o$) was achieved (1). Determination of resting tension (preload) at $L_o$ was accomplished as described previously (1). Length and force measurements were normalized as a fraction of $L_o$ and $P_o$, respectively.

Length changes were measured with an inductive transducer incorporated in a light lever assembly (Hugo Sachs Elektronik, Hugstetten, Germany). The original lever was replaced by a lighter aluminum-magnesium I-beam with movement restricted by upper and lower micrometer stops installed at the rear of the lever. The desired afterload was applied at a point giving the muscle a five-to-one mechanical advantage. The equivalent mass of the lever plus muscle connections was 1.07 g as determined by the method of Jewell (4). Because of the very slow shortening velocities of this tissue, neither the equivalent mass nor the electrical damping (30% at 10 Hz) of the displacement signal significantly affected the mechanical measurements. The output of the transducer was linear over the displacements studied (0-2.5 mm) and was accurate to ± 0.02 mm. A mechanical stop was positioned below the rear of the lever to avoid stretching the unstimulated muscle. A second stop controlled by a solenoid was located above the rear of the lever. Responses were isometric until this stop was suddenly lifted. The excursion of the upper stop was adjusted to limit muscle shortening to 30% of $L_o$ or less. Force was measured using four semiconductor strain gauges bonded to the lever beam in a bridge configuration so that two arms were in compression and two under tension (manufactured and installed by Kulite Semiconductor Products). The bridge was excited by a Brush transducer coupler. The output of the strain gauge was linear with the forces used (0-60 g). The compliance of the lever system under "isometric" conditions was 6.1 μm/g. Both force and length were monitored simultaneously, and the outputs were displayed on a fast rectilinear recorder (Brush model 220). The force-velocity measurements were performed by the quick-release method of Jewell and Wilkie (5). Potassium-induced responses produced about 20% more isometric tension than did electrical field stimulation; however, preliminary experiments revealed that the velocity of shortening under a finite afterload was greater when electrical stimulation was used. Similar findings have been previously reported (6). A supramaximal transverse electrical field stimulus (60 Hz, 20 v/cm peak to peak [2]) was applied, and the muscle was allowed to contract isometrically against the upper rear stop. After a delay of 5 or 10 seconds, the solenoid was activated, and the muscle was allowed to shorten isotonically after the quick release. The response consisted of a very rapid elastic recoil followed by a much slower shortening of the CE in which velocity was a function of the load (Fig. 1, bottom). The initial CE shortening velocity was estimated by drawing a tangent to the initial portion of the displacements vs. time recording following the elastic recoil (Fig. 1).

Stress-strain data for the SEC were obtained by quick release after the attainment of maximum isometric tension induced by high-potassium solutions. The magnitude of the rapid elastic recoil after quick release represents the change in length of the SEC from the maximally extended length to some lower value related to the afterload (5). Damped oscillations (Fig. 1) obscured the exact magnitude of the SEC recoil. However, the point of intersection of the tangent drawn to the CE shortening with the initial rapid upstroke of the shortening record (Fig. 1) provided a close estimate of the exact magnitude of the SEC recoil. However, the point of intersection of the tangent drawn to the CE shortening with the initial rapid upstroke of the shortening record (Fig. 1) provided a close estimate of the magnitude of the SEC recoil (5). The extension of the SEC with a given afterload was determined as the difference between the magnitude of the initial recoil under zero afterload and that obtained with a given afterload.

At the end of all experiments, the muscles were cut from between the clips, lightly blotted, and weighed. Cross-sectional area (cm²) was calculated by dividing the muscle weight by the product of its specific gravity (1.05 g/cm³) and $L_o$ (1). Oxygen diffusion at a PO₂ of 600 mm Hg did not limit contractile responses in these strips, which had a maximum thickness of 0.5 mm (1).
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FIGURE 2

Example of the force-velocity relationship for one muscle. Left: Solid circles represent the measured points. The curve is a hyperbola of the general form shown in the equation. Right: Regression line fitted to the data plotted in a linear form used to obtained the curve on the left. Values for Vmax, a/Po, and b were obtained from this plot.

Results

An example of the force-velocity relationship for a vascular strip is shown in Figure 2 (left). Actual measurements of force and velocity were normalized to P0 and L0, respectively, and are denoted by the solid circles. Velocity increased nonlinearly as the afterload was decreased. The solid line in the figure represents a segment of a rectangular hyperbola which conforms to the Hill equation (7) in the normalized form (equation on Fig. 2, left). The constants, a/Po and b, were determined by plotting (1 - P/P0) / V against P/P0 to obtain a straight line as indicated by the linearized form of the Hill equation (Fig. 2, right). The slope is 1/b and the intercept at the abscissa is -a/P0. The maximum velocity of shortening (Vmax) under conditions of zero afterload can be obtained from the ordinate intercept and equals P0b/a. For seven strips the following values (± SE) were obtained: Vmax = 0.12 ± 0.01 L0/sec, b = 0.020 ± 0.002 L0/sec, P0 = 1.80 ± 0.30 kg-wt/cm² (1.76 ± 0.29 × 10⁶ dynes/cm²), and a/P0 = 0.18 ± 0.02. The mean value of L0 in the strips used was 0.94 ± 0.03 cm.

Figure 3 (solid line) represents the average force-velocity curve determined from the mean values for the constants given above and plotted in absolute units of tension and velocity. The broken line in the same figure represents the power output obtained as the product of force and velocity. The units for power shown on the right ordinate are normalized to the cross-sectional area of the strips. Maximum power output calculated from the equation for the curve occurred at a load of 0.50 kg-wt/cm² or 0.28 P0.

The stress-strain curve for the SEC is shown in Figure 4 (left). The extension of the SEC is plotted as a percent of L0 against the afterload. The extension of the SEC at P0 for potassium-stimulated
strips (2.2 kg-wt/cm²) was 7.2 ± 0.5% L₀. Except at very low forces, the stress-strain curve can be described by a simple exponential. Average values for strain plotted as a function of log stress (Fig. 4, right) resulted in a straight line. The equation (Fig. 4, right) describing the stress-strain relationship was determined from the regression line fitted to the experimental data.

**Discussion**

The available force-velocity constants for mammalian smooth muscles are shown in Table 1. The $V_{\text{max}}$ of 0.12 L₀/sec found for the hog carotid artery is relatively low, particularly when it is compared with the results (Table 1) obtained at temperatures below 37°C. This finding is true even with respect to vascular smooth muscle. Hellstrand et al. (18) have shown that the rat portal vein shortens at 0.09 L₀/sec under an afterload of approximately 0.3 P₀; graphical extrapolation to zero afterload would yield a $V_{\text{max}}$ far exceeding that of the hog carotid strips. Laszt (19) has reported higher values for various vessels from the cow and the horse.

**TABLE 1**

<table>
<thead>
<tr>
<th>Tissue (source, temperature)</th>
<th>$P₀$ (kg-wt/cm²)</th>
<th>$V_{\text{max}}$ (L₀/sec)</th>
<th>$a/P₀$</th>
<th>$b$ (L₀/sec)</th>
<th>SEC (%L₀)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Carotid artery*</td>
<td>1.80</td>
<td>0.12</td>
<td>0.18</td>
<td>0.02</td>
<td>7.2</td>
</tr>
<tr>
<td>(hog, 37°C)</td>
<td></td>
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<td></td>
</tr>
<tr>
<td>Trachealis (8, 9)</td>
<td>1.17</td>
<td>0.17</td>
<td>0.21</td>
<td>0.04</td>
<td>7.5</td>
</tr>
<tr>
<td>(dog, 30°C)</td>
<td></td>
<td></td>
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<tr>
<td>Intestinal (10)</td>
<td>0.42</td>
<td>0.09</td>
<td>0.08</td>
<td>0.01</td>
<td>25</td>
</tr>
<tr>
<td>(cat, 30°C)</td>
<td></td>
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<tr>
<td>Taenia coli (11)</td>
<td>1.5</td>
<td>0.3</td>
<td>0.17</td>
<td>0.05</td>
<td></td>
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<tr>
<td>(guinea pig, 36°C)</td>
<td></td>
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<tr>
<td>Taenia coli (12, 13)</td>
<td>0.89</td>
<td>0.03</td>
<td>0.33</td>
<td>0.01</td>
<td>19-23</td>
</tr>
<tr>
<td>(rabbit, 22°C)</td>
<td></td>
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<tr>
<td>Uterus†</td>
<td>0.13</td>
<td>0.18</td>
<td>0.44</td>
<td>0.09</td>
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<td>(rabbit, 27°C)</td>
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<tr>
<td>Carotid artery (14)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>8.6-11</td>
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<td>(dog, 37°C, intact)</td>
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<td></td>
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<tr>
<td>Mesenteric artery (15)</td>
<td></td>
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<td></td>
<td>30</td>
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<tr>
<td>(calf, 37°C)</td>
<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Taenia coli (16, 17)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>12-15</td>
</tr>
<tr>
<td>(guinea pig, 37°C)</td>
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*Data from this study.
†Cited from reference 12.
although the shortening velocities for bovine and equine carotids (0.19 and 0.14 \(L_o/\text{sec}\), respectively) are similar to the value reported in the present paper. It is possible that smooth muscles from various sources exhibit a spectrum of maximum shortening velocities as is true for skeletal muscle (20).

The overall force-velocity relationship in the carotid strips (Fig. 3) is qualitatively similar to that of skeletal muscle, since it can be fitted by the hyperbolic relationship described by Hill (7). A discrepancy exists, however, between the individual shortening records seen with skeletal muscle (3, 5) and those obtained with the carotid strips (Fig. 1). In skeletal muscle, shortening velocity is relatively constant over a considerable distance. The vascular smooth muscle shows a continuously decreasing velocity with shortening. This behavior may be due to two factors acting in concert: (1) during shortening the load that the muscle must lift becomes progressively greater as the preload (10% of \(P_o\) at \(L_o\) [1]), which was initially born by the PEC, is transferred to the CE and becomes an afterload and (2) during shortening the CE moves down its length-tension curve with a progressive decrement in its force-generating capacity. For example, a 5% decrease in muscle length from \(L_w\) would add 8-9% of \(P_o\) to the effective afterload as determined from the passive and active length-tension curves for the carotid strips (1). An increase in afterload of this magnitude will lower shortening velocity proportionally when the strip has shortened to 0.95 \(L_w\), as determined from Figure 3. The estimated shortening velocities obtained by drawing tangents to the shortening records at a time corresponding to a length of 0.95 \(L_o\) were equal to the predicted values within experimental error.

Maximum tension development (\(P_o\)) is higher than that reported for other smooth muscles and is comparable to that in skeletal muscle (1-3 kg-wt/cm² [20]). However, a variety of factors including cell orientation, degree of activation, extracellular space, and perhaps temperature (12) complicate comparisons.

The estimates of series elasticity in smooth muscle vary considerably (Table 1). Our value of 7.2% of \(L_o\) is lower than the values generally reported, although it is more comparable to the value of 5-7% usually reported for mammalian skeletal muscle (20). Since maximum series compliance is a function of \(P_o\), our low value is remarkable in that it was obtained at higher tension than that generated by other mammalian smooth muscles. Several factors could lead to an overestimate of the smooth muscle SEC measured in vitro: (1) underestimation of \(L_w\), (2) underestimation of the preload at \(L_w\), so that the PEC contributes to the observed recoil in the quick-release experiments, and (3) dead tissue at the ends of the muscle. The variation in reported values may also reflect intrinsic properties of the different preparations.

The dynamic constants and the magnitude of the series compliance are similar in the carotid strips and canine trachealis smooth muscle (Table 1). The peak of the power output curve for the trachealis preparation occurs at 0.288 \(P_o\) (8) compared with 0.28 \(P_o\) obtained in the present study (Fig. 3), and both muscles exhibit a low resting tension at \(P_o\) (1, 8). These similarities may, at least in part, reflect the relative anatomical simplicity of both these parallel-fibered smooth muscle preparations.

Our studies represent an attempt to describe the mechanics of vascular smooth muscle in terms sufficiently general to allow comparison with other muscle types. Although there are important quantitative differences in the measured mechanical relationships, the qualitative similarities are impressive and are consistent with the three-component models proposed by Hill (7) and others for skeletal muscle. This finding implies that the contractile system of smooth muscle may share common features with that of striated muscle.

Acknowledgment

Special thanks are due to Mrs. Mildred Smythers for her technical assistance.

References

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Circ Res. 1974;34:461-466
doi: 10.1161/01.RES.34.4.461

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