Effect of Torso Resistivity Variation on the Electrocardiograms of Children, Using a Grid Lead System

By Eugene J. Fischmann, M.D., Mark R. Barber, Ph.D., and Herbert H. Lehner

SUMMARY
Most clinical ECG leads determine cardiac currents from a few surface potentials, without quantitating torso geometry and structure, by empirical-intuitive methods. The present report is part of a broader study asking whether the known limitations of ECG can be reduced by multi-electrode grid leads which sample extensively, measure torso geometry and structure, and use clearly defined biomathematics.

One torso characteristic not measured in clinical ECG is torso resistivity \( \rho \), while past studies of the cardiac dipole moment replaced measurement of \( \rho \) by assuming that it is 480 ohms-cm. In essence this amounts to using Ohm's law: Current = voltage/resistance without consideration of the resistance term.

The present work attempts to measure each patient's \( \rho \) as part of the ECG recording procedure, by one of two methods: Dipole moment and \( \rho \) are separately determined and the measured dipole moment (Mm) is later corrected by computer, or the ECG recorder is calibrated for the patient's \( \rho \), resulting in an ECG directly read as corrected dipole moment, ma-cm.

The effect of \( \rho \) variation on the ECG was assessed in 51 children by comparing instantaneous and maximal Mm and its components when \( \rho \) was an arbitrary 480 ohms-cm (Mm 480) or individually measured (Mmi). Measured \( \rho \) was less than 480 ohms-cm in all subjects and decreased with age. The differences between Mm 480 and Mmi were typically half to a third of the mean Mm 480 and about half of the Mm 96 percentile. Correction of the ECG for individual \( \rho \) should result in revaluation of estimates of the heart's total force, as did area correction in previous reports of this series.

Additional Indexing Words:
Heart dipole moment Individual lead calibration Multi-electrode leads Surface potential integrating leads

Presently used clinical ECG and VCG methods evaluate cardiac currents from surface potentials without measuring torso resistivity. The nonmathematical reader may grasp the possible significance of this by considering that ECG is in essence an extended application of Ohm's law. As current flowing in a wire can be determined from the relationship between the potentials on and the resistivity of the wire, so may we attempt to estimate currents within the heart from torso resistivity and torso surface potentials. The conventional ECG uses "Ohm's law, \( I = V/R \), without the resistance term R." In view of the

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Supported by grants from the Washington Heart Association and the Ontario Heart Foundation.


Circulation, Volume XLII, July 1970

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Received December 29, 1969; revision accepted for publication March 24, 1970.
known limitations of conventional ECG, it seems justifiable to ask how far determining torso resistivity as well as surface voltage may change and possibly improve ECG estimates of cardiac currents. The parallel grid lead system\textsuperscript{1} seems suited to test this proposition, since it uses a simple relationship of surface potential, torso area, and torso resistivity in determining the heart’s total force expressed as dipole moment (equation 5) and as shown by Ellison and his associates\textsuperscript{2–4} readily lends itself for clinical studies.

Methods

The present grid differs from previous versions\textsuperscript{1, 2} in the following: In our own and other past work with integrating systems, constant mean torso resistivity of 480 ohms-cm, was assumed,\textsuperscript{1–5} a probable source of error since individual mean torso resistivity varies from 150 to 650 ohms-cm.\textsuperscript{6–9} The present grid system was therefore designed to determine individual mean torso resistivity as part of the ECG recording procedure and to correct the ECG for torso resistivity. Grid design, including electrode numbers, was previously based on model studies. It is now modified to accord with surface potential map\textsuperscript{10} and patient\textsuperscript{11} experience. Further differences in design detail are as follows:

Mechanics and Accessory Circuitry

Since Ellison and co-workers\textsuperscript{2} described in detail a recent grid system prototype, we shall state only the differences between the latter and the present system: It was shown by Brody and Arzbache\textsuperscript{12, 13} and by Fischmann and Elliott\textsuperscript{14, 15} that a sufficient number of electrodes is as essential in the X as in the Z lead. In the preceding grid prototype we replaced the X grid by an approximation; the present device has complete X and Z grid-pairs. All information needed for torso area calibration is now entered by simply setting three rotary switches corresponding to the three (X, Y, and Z) diameters of sole operation required to calibrate the output of the system for projected torso area. The two-position switches on the reader’s left reduce the number of electrodes in contact without changing grid area for studies on the effect of electrode number on dipole moment. The terminal pegboard shown allows separate access to each electrode for surface potential mapping and the use of selected combinations. The torso resistivity measuring and calibration controls are on the reader’s right.

(Lower) System in place (patient aged 12 years).

\textit{Circulation, Volume XLII, July 1970}. 

\begin{figure}
\centering
\includegraphics[width=\textwidth]{figure1.png}
\caption{Some features of the grid system not present in earlier prototypes.}
\end{figure}

\begin{itemize}
\item \textbf{(Top)} The device now has transverse as well as sagittal electrode arrays. The anterior probes are no longer spring loaded. A thick double platform with knife-edged holes is an attempt to decrease the effect of rod friction. The effective probe weight \( W_{eff} \) is related to the actual weight by the following formula in which \( d \) is the chest to platform distance, \( t \) is platform thickness, \( \delta \) coefficient of friction and \( \theta \) the torso slope angle.

\[ W_{eff} = \frac{W}{(2d/t) \delta \tan \theta + 1} \]

\item \textbf{(Middle)} The three rotary switches can be set to the X, Y, Z diameters of the torso: This is now the...
the torso. Electrodes on the left and right arms can be connected into the X-grid outputs.\textsuperscript{16–18} This effectively extends the X-grid upward and thereby provides a more uniform sensitivity to dipole layers high in the chest. Terminal board access to each of the 275 electrodes (fig. 1) allows use of the grids for surface potential mapping, the selection of desired electrode combinations, or the attachment of logic circuits not at present included as foreseen by Helm and Chou.\textsuperscript{19} Switching to reduce the number of the X and Z electrodes without change in total grid areas is provided in the grid control box and was used in studies concerning adequate electrode numbers for dipole determination.\textsuperscript{10} The probes of the anterior grid are no longer spring loaded but are kept in position by their own weight. To diminish friction between the probes and the platforms, double platforms with knife-edge holes are used.

**Output**

Connected to an ECG recorder, the system produces three scalar orthogonal ECG leads, X, Y and Z. The leads can be directly read as orthogonal dipole moment components. If conventional ECG calibration is used, each millivolt of deflection is equivalent to between 1 and 4 milliamperes-cm (ma-cm) of component dipole moment. Planar and axial projections, areas, and loops can be derived as with other lead systems.

**Calibration of the Grid Electrocardiogram for Individual Mean Torso Resistivity**

**Subthreshold Measuring Current**

To measure resistivity, it is necessary to inject current into the human torso. It is obviously desirable to keep this at a level that does not produce cardiac contraction, let alone ventricular fibrillation. The introduction of cardiac pacers has led to extensive investigation of the heart's stimulation threshold as summarized in references 20 and 21.

The conductivity measuring apparatus housed in the control box of the grid system (fig. 1) delivers a square wave of 4.3 volts peak to peak or 2.15 volts RMS. A current of 60 microamperes peak to peak flows when 1 mv appears across the 17 ohm resistor in the equipment. The RMS value of this current is 30 microamperes. The power dissipated in the body is therefore $2.1 \times 30 = 63$ microwatts. If the oscillator yields a square wave of 100 Hz and five complete cycles are used to perform each resistivity measurement, then the energy dissipated in the body will be $E = 63 \times 5/100 = 3.1$ microjoules. This energy level is one seventh of the threshold energy of 20 microjoules\textsuperscript{20, 21} for epicardial pacing electrodes.

The safety factor of seven is augmented by the use in the resistivity measuring setup of an approximately tenfold greater electrode area and increased heart electrode distance, both known to increase the pacing threshold, and by a probable further factor of approximately 10, since in experiments on dogs, only about 10\% of longitudinally applied head-foot current passes through the heart.\textsuperscript{22} The energy needed to pace the heart is below the fibrillation threshold by a factor of 4 to 10.\textsuperscript{20, 21} At energy levels of 3.1 microjoules patients show no sensory awareness of the current, even if their attention is drawn to the moment of its application.

It seems reasonable to assume that this energy level applied to the skin of the leg and forehead with relatively large electrodes is well below the cardiac pacing threshold. That it is possible to measure torso resistivity safely, with much higher energies than those used by us, is shown by numerous studies.\textsuperscript{6–9}

**The Square-Wave Generator**

The use of square waves overcomes stray capacitance effects which appear as an initial spike. This is ignored. Polarization effects produce plateau slopes. The quality of electrode-skin contact should be good enough to give a square wave having a slope of 30° or less. True voltage is measured by measuring the up or downstroke before polarization effects have occurred. While using sine waves, it is not possible to eliminate capacitance and polarization so that increasing frequencies lead to errors of increasing magnitude by capacitance effects which cannot be separated from the desired wave form. A similar effect occurs for polarization at lower frequencies. Thus, when square waves are used, the choice of wave frequency is less important.

**Calibration of the Recorded ECG for Torso Resistivity Using a Computer**

A 100-Hz square-wave generator within the control box (fig. 1) injects 60 microamperes peak to peak alternating current, through the head-foot lead Y. Two electrodes of the anterior grid separated 10.7 cm (or 3-unit spacings) in the longitudinal torso axis can now function as "voltage" electrodes. The patient should have only one ground connection through the right leg. As shown previously, the equation relating the voltage $V_x$ appearing between head and foot with the Y-directed moment $(\mu_Y)$ of a dipole within the heart is:

$$\mu_Y = V_x \frac{A_{yz}}{\rho} \quad (1)$$

*In mathematical models $\mu$ will be used for dipole moment. Reciprocity allows interchange of the current and voltage electrodes.
where $A_{xx}$ is the torso cross-sectional area in the region of the heart (cm$^2$). If, instead of the heart dipole, an artificial dipole, excited with a square-wave generator, is placed on the chest, then the same equation will apply. Let the current into this chest dipole be measured by dividing the voltage $V_1$ across a small series resistor by its value $R = 17.2$ ohms. That is:

$$\mu_y = (V_1/R)d$$  \hspace{1cm} (2)

where $d$ is the spacing of the Y-directed current electrode pair (cm). Hence from equations 1 and 2

$$V_1/R)d = V_y A_{xy}/\rho \text{ or }$$

$$\rho_y = R V_y A_{xx}/dV_1 \text{ ohms-cm.}$$  \hspace{1cm} (3)

Equation 3 requires measuring three quantities, $V_1$, $V_y$, and $A_{xx}$. $R$ and $d$ are fixed by the design of the measuring apparatus: In the present grid, for example, $R = 17.2$ ohms and $d = 10.7$ cm. Hence

$$\rho_y = \frac{1.6 V_y A_{xx}}{V_1} \text{ ohms-cm.}$$  \hspace{1cm} (4)

This value of $\rho$ can be used in equation 5a and similarly in equations 5b and 5c, to yield the three components of the heart’s measured dipole moment (Mm).

The grid lead system is a relatively simple special purpose computer for determining three orthogonal components of the heart’s total dipole moment by solving the three equations:

$$M_{m_x} = V_x A_{ys}/\rho.$$  \hspace{1cm} (5a)

$$M_{m_y} = V_y A_{xv}/\rho.$$  \hspace{1cm} (5b)

$$M_{m_z} = V_z A_{xy}/\rho.$$  \hspace{1cm} (5c)

In equation 5a, $M_{m_x}$ is the transverse component of the heart’s total dipole moment measured with the grid, $V_x$ transverse grid-lead voltage $A_{ys}$ lateral grid “area,” and $\rho$ torso resistivity; the sagittal Z and longitudinal Y dipole components are determined similarly in equations 5b and 5c.

**Direct Recording of Dipole Moment (Mm) Calibrated for Torso Resistivity**

The present grid system contains logic for torso area correction. Torso diameters in the X, Y, and Z directions are entered into the control box with rotating voltage dividers. The apparatus will then yield the three components of Mm according to equations 5a, 5b, and 5c. When constructing the grid system, it is convenient to have the maximal projected grid area in the three equations related in a simple manner, for example in the present grid

$$A_{ys} = A_{xx} = \frac{1}{2} A_{xy}.$$  

The Y channel is then calibrated for torso resistivity as follows: The square wave of current representing the artificial chest dipole strength divided by its electrode separation is displayed on the recorder (screen or strip) and its amplitude is measured. The corresponding square wave measured between head and foot is displayed on the Y channel, and the Y-amplifier gain is adjusted so that the square-wave excursion again corresponds to dipole magnitude. If the X channel is then adjusted to equal Y, and the Z channel to have double this gain, the grid system will be calibrated with compensation for “torso resistivity.” The graphic display will yield the numerical value of the three components of Mm in ohm-cm units.

**Effect of Change from Arbitrary to Individually Measured Torso Resistivity on Measured Moment**

To avoid dependence of the data on the assumption of a normal distribution, the maxima and 96 percentiles are also used in evaluating the findings, in addition to parameters of a hypothetical Gaussian distribution.

The ages of the 51 children investigated in the present study ranged from 5 years and 4 months to 15 years and 1 month; 27 were male. All had normal cardiovascular and respiratory systems and were admitted to the Hospital for Sick Children for minor ailments not affecting these systems or were the healthy offspring of University staff. The resistivities encountered were all under the previously assumed arbitrary amount of 480 ohm-cm.

The range defined by the 4 and 96 percentiles was 280 to 410 ohm-cm, with a mean of 350 and standard deviation of 46 ohm-cm (SE, 6 ohm-cm). Thus in this series the range of variation was not as wide as in previously described adult groups where it ranged 160 to 650 ohm-cm, but accorded well with the values found by Gamboa and Adair in children of comparable age.

The data in tables 1 to 4 show that even this lesser variation in assumed “mean” resistivity induces substantial departures of the measured dipole moment (Mm) from the values obtained when resistivity is arbitrarily set at 480 ohm-cm in all patients. Commencing 5 msec after QRS onset at 11 instants separated by 5 msec, the following measurements were
obtained in each of the 51 children: (1) Mm 480, measured dipole moment, when mean torso resistivity is assumed to be 480 ohms-cm; (2) Mmi, dipole moment when torso resistivity is individually measured; (3) the differences between Mm 480 and Mmi; (4) X, Y, and Z components of Mm 480 and Mmi; (5) the maximum Mm 480 and Mmi, and the difference between the two, in each subject; (6) the maximum X, Y, and Z components of Mm 480 and Mmi, and the differences between these in each subject. Since the measured mean resistivity was less than 480 ohms-cm in every child in this series, Mmi was greater in every instance than Mm 480, except early in QRS where Mm was small and the differences were within the ranges of measurement error and noise (table 1). As expected the greatest Mm 480 to Mmi changes occurred in the midportion of QRS.

Table 1 shows the distribution of Mm 480 and Mmi in the 51 children, at each of the 11 QRS instants. Mean and 96 percentile maxima occur at 30 and 35 msec toward the center of QRS; mean Mm 480 and Mmi maxima are 0.36, and 0.50, Mm 480 and Mmi 96 percentile maxima 0.79 and 1.21 ma-cm, respectively; the greatest mean difference between Mm 480 and Mmi is 0.17, that of the 96 percentile 0.55 ma-cm, both at 35 msec. Thus, in this form of data presentation the maximum change in the mean instantaneous dipole moment, induced by changing from arbitrary to measured \( \rho \) is approximately one half of the greatest Mm 480 and one third of the greatest Mmi in the group. The greatest induced change in the Mm 96 percentile exceeds both the Mm 480 and Mmi maxima and is approximately two thirds of the Mm 480, and approaches half of the Mmi, 96 percentile maxima.

Instead of pooled data as in table 1, table 2 presents the distribution of Mm 480 to Mmi

### Table 1

**Measured Instantaneous QRS Dipole Moments in 51 Children**

<table>
<thead>
<tr>
<th>QRS intervals (msec)</th>
<th>Mm 480 and Mmi (ma-cm)</th>
<th>Mean</th>
<th>sd</th>
<th>se</th>
<th>Max</th>
<th>96%</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
<td>0.08*</td>
<td>0.04</td>
<td>0.01</td>
<td>0.20</td>
<td>0.15</td>
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</tr>
<tr>
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<tr>
<td>15</td>
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<td>0.59</td>
<td>0.42</td>
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<tr>
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<td>0.36</td>
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</tr>
<tr>
<td>20</td>
<td>0.23</td>
<td>0.12</td>
<td>0.02</td>
<td>0.69</td>
<td>0.45</td>
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<tr>
<td></td>
<td>0.25</td>
<td>0.17</td>
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<td>0.89</td>
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<td>1.49</td>
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<td>0.02</td>
<td>0.69</td>
<td>0.51</td>
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</tr>
</tbody>
</table>

*In each pair, Mm 480 is first; Mmi second.

Abbreviations: Mm 480 = dipole moment when mean resistivity is assumed to be 480 ohms-cm; Mmi = dipole moment when resistivity is individually determined.

### Table 2

**Differences in Measured Instantaneous Dipole Moment (Mm) in 51 Children, Caused by Changing from an Arbitrary Mean Torso Resistivity of 480 ohms-cm to Estimated Individual Resistivity (Mmi in ma-cm)**

<table>
<thead>
<tr>
<th>Interval from QRS onset (msec)</th>
<th>Instantaneous Mmi — instantaneous Mm 480</th>
<th>Mean</th>
<th>sd</th>
<th>se</th>
<th>96%</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
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<td>0.15</td>
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</tr>
</tbody>
</table>

Abbreviations: Mmi and Mm 480 = dipole moment with torso resistivities individually estimated, and arbitrarily fixed, respectively. Mmi, Mm 480, and the differences between the two values were determined in each patient at 5-msec intervals, commencing 5 msec after QRS onset. Mmi was greater than Mm 480 in every child in the series. The table therefore shows the distribution of 51 values of Mmi — Mm 480 at each of 11 consecutive instants of QRS.
changes, induced by changing from arbitrary to measured $\rho$ obtained individually in each of the 51 subjects. Here again large differences are seen; for example, the maximum mean change, 0.15 ma-cm, is at 35 msec, the greatest 96 percentile change, 0.36 ma-cm, is at 40 msec and the maximum mean induced change approaches one half of the mean Mm 480 and two thirds of the mean Mmi in table 1.

The data in table 3 were obtained by registering 11 consecutive instantaneous X, Y, and Z components of Mmi and Mm 480 at 5-msec intervals commencing 5 msec after QRS onset. The maxima of the six 11-measurement sets, and the maximum difference between Mmi and Mm 480 was found for each patient. The table shows the distributions of these maxima in the 51 children. The induced differences are again substantial. Thus, the 96 percentile of the differences in Mm$_x$ maxima, induced by the change from assumed to measured $\rho$, approaches the mean of the Mm 480 maxima and two thirds of the corresponding mean Mmi maxima. In Mm$_y$, the 96 percentile difference 0.38 ma-cm exceeds the Mm 480, and is 79% of the Mmi, mean maximum. In Mm$_z$, the mean Mm 480 to Mmi difference of 0.08 ma-cm is about half of the Mm 480 mean maximum and one third of the Mmi mean maximum, whereas the 96 percentile 0.16 approaches, and is 59% respectively of, these means.

Table 4 shows the variation of the greatest instantaneous Mm measurements obtained separately in each of the 51 children. Thus, for instance, 0.50 ma-cm, the 96th percentile difference between Mm 480 and Mmi, is greater than mean Mm 480, approaches Mm 450, and is about half of Mmi at the 96th percentile. Figure 2 shows computer plots of Mm 480 and Mmi throughout QRS in two typical patients.

**Discussion and Conclusions**

In ECG lead research aimed at reducing the still impressive diagnostic error of clinical electrocardiography, it is convenient to treat the ECG as a single communication system: Disease encodes a message on the cardiac generator by changing it in more or less specific ways. The message also appears as potential variation on the torso surface and as a graphic display on the recorder. Reading the ECG is decoding the message. Ideal communication systems (1) do not change the message in transit and (2) deliver it in easily decodable form. ECG is not an ideal system for its diagnostic error suggests that the message is distorted and it is not easily decoded as is shown by the vast and still incomplete 50-year research effort that went

### Table 3

**Differences in the Maximum Instantaneous QRS Amplitude of the X, Y, and Z Components of the Measured Dipole Moment (Mm), Caused by Changing from Arbitrary Torso Resistivity of 480 ohms-cm to Individually Estimated Torso Resistivity (Mm in Ma-cm, $\rho$ = Torso Resistivity)**

<table>
<thead>
<tr>
<th>Component</th>
<th>Mean</th>
<th>SD</th>
<th>SE</th>
<th>96%</th>
</tr>
</thead>
<tbody>
<tr>
<td>Orthogonal components of Mm when $\rho$ is assumed 480 ohms cm (Mm 480)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>X</td>
<td>0.31</td>
<td>0.14</td>
<td>0.02</td>
<td>0.58</td>
</tr>
<tr>
<td>Y</td>
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<td>0.18</td>
<td>0.03</td>
<td>0.67</td>
</tr>
<tr>
<td>Z</td>
<td>0.19</td>
<td>0.08</td>
<td>0.01</td>
<td>0.31</td>
</tr>
<tr>
<td>Orthogonal components of Mm when $\rho$ is measured (Mmi)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>X</td>
<td>0.43</td>
<td>0.18</td>
<td>0.03</td>
<td>0.77</td>
</tr>
<tr>
<td>Y</td>
<td>0.48</td>
<td>0.27</td>
<td>0.04</td>
<td>0.99</td>
</tr>
<tr>
<td>Z</td>
<td>0.27</td>
<td>0.11</td>
<td>0.02</td>
<td>0.48</td>
</tr>
<tr>
<td>Difference due to change from assumed to measured $\rho$</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>X</td>
<td>0.13</td>
<td>0.08</td>
<td>0.01</td>
<td>0.28</td>
</tr>
<tr>
<td>Y</td>
<td>0.15</td>
<td>0.11</td>
<td>0.02</td>
<td>0.38</td>
</tr>
<tr>
<td>Z</td>
<td>0.08</td>
<td>0.04</td>
<td>0.01</td>
<td>0.16</td>
</tr>
</tbody>
</table>

### Table 4

**Differences in Maximum Measured QRS Dipole Moment (Mm) in 51 Children, Caused by Changing from an Arbitrary Mean Torso Resistivity of 480 ohms-cm to Individually Estimated Resistivity (Mm in ma-cm)**

<table>
<thead>
<tr>
<th></th>
<th>Mean</th>
<th>SD</th>
<th>SE</th>
<th>96%</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mm 480</td>
<td>0.44</td>
<td>0.16</td>
<td>0.02</td>
<td>0.71</td>
</tr>
<tr>
<td>Mmi</td>
<td>0.63</td>
<td>0.24</td>
<td>0.03</td>
<td>0.96</td>
</tr>
<tr>
<td>Mm 480-Mmi</td>
<td>0.19</td>
<td>0.12</td>
<td>0.02</td>
<td>0.50</td>
</tr>
</tbody>
</table>

Abbreviations: Mm 480 = measured moment when $\rho$ is 480; Mmi = measured moment when $\rho$ is individually estimated. Difference when Mm 480 and Mmi were separately obtained in each child; the last row shows the distribution of the differences in the group.

*Circulation, Volume XLII, July 1970*
Figure 2

Computer plot of grid outputs, in two subjects. C assumes constant mean torso resistivity of 480 ohms-cm, and V uses the subject's individually measured resistivity. Abscissa, time msec × 10⁻¹; ordinate, dipole moment ma-cm × 10.

into ECG interpretation and by ECG reader variation in interpreting ECG records.

A large section of the ECG communication system lies within the human torso through which the message must pass on its way to the surface. This, the given, immutable, and in geometry and structure unknown, section of the system is the major source of its vulnerability. It is one of the main functions of the designed section of the communication pathway which includes the ECG lead, to supply circuitry to correct the distortion in the torso.

The work here reported is part of a continued effort to find ways of correcting ECG distortion due to the geometry and resistivity of the torso and through individual variation of these two characteristics. Past studies quoted elsewhere² have shown that the variable torso geometry is an important source of distortion. The grid lead system has made it possible to correct for geometry expressed as three orthogonal-plane torso projections. Ellison and his associates²–⁴ have shown that a lead system able to make this correction produces ECG measurements that differ significantly from measurements obtained with leads that do not make the correction. In a second step these authors have shown that the new corrected values yield clinically useful

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improvements in previous heart-ECG correlations and also allow new, previously not available correlations.2-4

The work here reported attempted to assess the extent to which interindividual changes in yet another variable, namely torso resistivity, affect ECG measurements. It appears that corrections for torso resistivity result in ECG data that differ substantially from data that do not take torso resistivity into account. Ignoring interindividual variation in torso resistivity results in an apparent error of one half to two thirds of the measurement range. The magnitude of the error makes it seem reasonable to progress to the next step: namely, investigating the possibility of clinical gain by using ECG measurements corrected for both torso dimensions and torso resistivity.

Sources of Error

One group of currently employed orthogonal leads, including the Frank, derives the cardiac generator from surface voltage measurements, modified by scale factors derived from homogeneous torso models. This involves assuming identity of the model and all human torsos. The grid does not depend on the model-human analogy assumption, since it derives the generator from surface voltages without the use of scale factors.

Ideal ECG leads should adequately deal with the following variables: surface area, shape, and mean and local resistivity of the torso; size, shape, location, and mean and local resistivity of the heart; intracardiac generator distribution; skin resistance; and surface electrode position. Recent work has shown both the importance and feasibility of controlling some of these variables.2-4, 28, 24 Methods using parallel grids appear to have achieved partial control. The assumption implied in conventional ECG leads, that surface area and configuration are of limited relevance, need not be made when using the grid, since the grid output is calibrated for area. As the grid deals with projections of the torso surface, it is at least partly independent of surface configuration. The single dipole assumption is not needed since the surface voltage integral recorded by the grid is the sum of dipole moments within the torso irrespective of their number or position.19 (The single dipole assumption is reintroduced, however, when using grid leads for vectorcardiography.) Positional variation of the heart and the position and distribution of the generator within it and changes in electrode position should have a diminished effect on grid leads, since moving the grid in relation to the patient or heart does no more than relate the patient or heart to another set of similarly weighted and therefore equivalent set of electrodes.

Resistivity variation from subject to subject and from point to point within each torso is a serious bar to ECG quantitation. Methods determining cardiac force as total dipole moment from surface potential integration, including the grid, have moved toward solving the problem of torso resistivity in the following steps: Gabor and Nelson,25 Barber and Fischmann,1 Nelson,5 and Ellison and associates2-4 have assumed a constant mean torso resistivity of 480 ohms-cm. The grid prototype described in the present report replaces the constant resistivity assumption by calibration for individually measured mean torso resistivity. Elimination of two assumptions, that mean resistivity is adequately represented by a single resistivity measurement and that torso resistivities in the X, Y, and Z directions are identical, should be the object of further study. A method of using multiple measurements of resistivity to approximate further "mean torso resistivity" is being developed.

While the present work represents an attempt to cope with inter-subject and time-dependent intra-subject variation of torso resistivity, the inhomogeneous resistivity of the torso, which is among the most serious obstacles to ECG quantitation, remains uncontrolled. This may prove an intrinsic and therefore unsurmountable difficulty, since total dipole moment determination as integrated surface potential depends on the assumption of a homogeneous field.

Acknowledgment

We wish to thank Prof. A. A. Bishop, Syracuse University, Dr. Robert A. Dalton, Corning Glass
References

22. Kouwenhoven WB, Hooker DR, Langworthy OR: The current flowing through the heart under conditions of electric shock. Amer J Physiol 100: 344, 1932
Effect of Torso Resistivity Variation on the Electrocardiograms of Children, Using a Grid Lead System
EUGENE J. FISCHMANN, MARK R. BARBER and HERBERT H. LEHNER

_Circulation_. 1970;42:171-179
doi: 10.1161/01.CIR.42.1.171

_Circulation_ is published by the American Heart Association, 7272 Greenville Avenue, Dallas, TX 75231
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Print ISSN: 0009-7322. Online ISSN: 1524-4539

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