Physiological Noise in Oxygenation-Sensitive Magnetic Resonance Imaging

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The physiological noise in the resting brain, which arises from fluctuations in metabolic-linked brain physiology and subtle brain pulsations, was investigated in six healthy volunteers using oxygenation-sensitive dual-echo spiral MRI at 3.0 T. In contrast to the system and thermal noise, the physiological noise demonstrates a signal strength dependency and, unique to the metabolic-linked noise, an echo-time dependency. Variations of the MR signal strength by changing the flip angle and echo time allowed separation of the different noise components and revealed that the physiological noise at 3.0 T (1) exceeds other noise sources and (2) is significantly greater in cortical gray matter than in white matter regions. The SNR in oxygenation-sensitive MRI is predicted to saturate at higher fields, suggesting that noise measurements of the resting brain at 3.0 T and higher may provide a sensitive probe of functional information. Magn Reson Med 46: 631–637, 2001. © 2001 Wiley-Liss, Inc.

Key words: neuroimaging; spiral scan; magnetic field strength; SNR; physiological noise

Several biophysical models (1,2) and recent investigations of the field dependency in MRI (3–8) strongly suggest improvements in the signal-to-noise ratio (SNR) and the contrast-to-noise ratio (CNR) at higher magnetic fields. In a recent work (8), however, we have shown that the physiological noise demonstrates an MR signal strength dependency and exceeds the thermal noise and scanner noise at 3.0 T. Consequently, the physiological noise counteracts the gain in SNR at higher fields. In the present study, a model for the intrinsic noise in oxygenation-sensitive MRI of the human brain is developed. In corresponding experiments the MR-signal strength was modulated by variations in the flip angle (α) and echo-time (TE) to separate individual noise contributions in the resting brain. Results were compared with the predicted noise contributions and corresponding implications on neuroimaging modalities are discussed.

THEORY

The signal and the intrinsic noise in high-field MRI have been shown to be quadratic and linear in \( B_0 \), respectively, resulting in an SNR proportional to the strength of the magnetic field (3,4). Whereas these earlier studies characterize the dominant noise by thermally generated random noise from the subject and scanner electronics and assume that coil losses are negligible (4), we expand this model by considering physiological noise as a further significant contribution to the total image noise:

\[
\sigma = \sqrt{\sigma_T^2 + \sigma_S^2 + \sigma_N^2}.
\]

Here, \( \sigma_T \) is the thermal noise from the subject and scanner electronics and \( \sigma_S \) describes other system noise that includes drift and imperfections in RF, gradient, and shim subsystems. \( \sigma_N \), the square-law sum of \( \sigma_T \) and \( \sigma_S \), is considered the raw noise and has been shown to be proportional to \( B_0 \) (4) but independent of the MR-signal strength. The term \( \sigma_p \) in Eq. [1] describes the physiological noise, which arises from fluctuations in the basal cerebral metabolism (CMRO2), blood flow (CBF), and blood volume (CBV), but also from cardiac and respiratory functions that cause quasiperiodic oscillations in the vascular system (9–11), motion from subtle brain pulsatility (12), and magnetic field modulations. In contrast to the raw noise, the herein discussed physiological noise is signal-dependent.

In a more detailed consideration, we further distinguish between two different physiological noise terms, \( \sigma_B \) and \( \sigma_{NB} \). The first contribution describes fluctuations in the transverse relaxation rate \( R_2^* \) that gives rise to small changes in the \( T_2^* \)-weighted signal, \( \Delta S \). These fluctuations, which are closely linked to brain physiology, are caused by the same mechanism that results in activation-induced signal changes in BOLD imaging and can be described by \( \sigma_B = c_1 \cdot \Delta S \), where \( c_1 \) is a constant. Because of that analogy and with \( dS/dR_2^* = -TE \cdot S_0 \cdot \exp(-TE \cdot R_2^*) \), \( \sigma_B \) can be expressed as:

\[
\sigma_B = c_1 \cdot S \cdot \Delta R_2^* \cdot TE,
\]

where \( \Delta R_2^* \) represents the physiological fluctuation in \( R_2^* \). Note the TE-dependency in \( \sigma_B \) and that \( \sigma_B \) demonstrates a maximum at \( 1/R_2^* \). The second contribution arises from image-to-image signal fluctuations due to cardiac and respiratory functions and scanner imperfections that demonstrate no TE-dependence. We can express the second contribution as \( \sigma_{NB} = c_2 \cdot S \), where \( c_2 \) is a constant. Thus, the overall physiological noise

\[
\sigma_p = \sqrt{\sigma_B^2 + \sigma_{NB}^2}
\]

is signal-dependent, as both \( \sigma_p \) and \( \sigma_{NB} \) are proportional to the signal. Because \( \sigma_p \) increases at the same rate as the
signal \( S \), the achievable SNR at high signal strength and high magnetic fields reaches an asymptotic limit. From Eq. [8], the SNR \( = S/\sigma \) reaches can be expressed as (see also Ref. 4)

\[
\text{SNR} = \frac{\text{SNR}_0}{\sqrt{1 + \left( \frac{c_1 \cdot \Delta F^2 \cdot \text{TE}^2 + c_2}{\lambda} \right) \cdot \text{SNR}_0}}, \tag{4}
\]

with \( \text{SNR}_0 = S/\sigma_0 \) and \( \sigma_p = \lambda \cdot S \). In this model, \( \lambda \) demonstrates a physical measure of the SNR-degradation by signal-dependent fluctuations, such that if \( \lambda = 0 \) the SNR reverts to the thermal signal-to-noise ratio, \( \text{SNR}_0 \). \( \lambda \), \( \sigma_0 \), and \( \sigma_p \) are experimentally accessible by dynamic changes of the MR signal strength. Modulations of the signal, however, can be performed by changing the sequence parameter flip angle (as in this work), the echo time, or e.g., the voxel size. According to Eq. [4], changing TE can be used to distinguish the physiological noise contributions \( \sigma_0 \) and \( \sigma_{NB} \).

Fluctuations from system imperfections, e.g., in the RF and gradient subsystems, may exhibit an extremely complex behavior and may contribute to both the raw noise \( \sigma_0 \) and the signal-dependent physiological noise term \( \sigma_p \). However, on a well-adjusted system the expected contribution from these sources to the total image noise is relatively small. Furthermore, a recent investigation (13) of image-to-image fluctuations from the disturbance of steady-state free precession (SSFP) has demonstrated that such noise components may evolve in imaging procedures for which \( TR < T_2 \). Because we use a \( TR = 3.0 \text{ sec} \) in the current investigation, this noise source is irrelevant.

**METHODS**

**Imaging and Scaling of Sequence Parameters**

All experiments were conducted at 3.0 T (GE LX, rev. 8.25, General Electric Medical Systems, Milwaukee, WI, USA) using a custom-built transmit-receive imaging headcoil (14). Oxygenation-sensitive MRI acquisition was based on a gradient echo version of a single shot spiral sequence (15). To improve efficiency we used a dual-echo sequence. Acquisitions of both the first and the second echo started at the \( k \)-space origin and spiraled out to the maximum radius desired. A spectral-spatial RF-pulse was used to excite only water spins in the spatial slice of interest (16). After prescription of four contiguous, oblique slices (4 mm thickness, 0.5 mm gap) through the calcaneus fissure the magnetic field homogeneity in the respective volume-of-interest (VOI) was improved by an automatic second-order shim correction (17), typically resulting in root mean square (RMS) fluctuations smaller than 16 Hz in the depicted VOI. For this study, we used a matrix size of \( 64 \times 64 \) over a field-of-view (FoV) of \( 220 \times 220 \text{ mm}^2 \). This corresponds to a single-shot spiral readout window of 19.1 ms. A minimized sensitivity to inflow effects was accomplished by means of a long repetition time \( (TR = 3.0 \text{ sec}) \) in all experiments. For retrospective correction of physiological motion effects, respiration and cardiac functions were monitored during the experiments using a photo-plethysmograph and a pneumatic belt, respectively (18). This image-based pixel-wise correction scheme has been shown to provide substantial reduction of the noise energy from cardiac and respiratory functions even when the noise is temporally aliased by undersampling, as in the present study \( (TR = 3.0 \text{ sec}) \). Image reconstruction was performed off-line using a gridding algorithm (19). A retrospective slice-by-slice first-order shim correction was performed using a \( B_0 \)-field map calculated from the first two frames with different echo-times (ATE = 2 ms) (20). Conventional \( T_1 \)-weighted gradient echo images \( (TR/TE/\text{flip} = 70/5/60^\circ) \) were obtained for anatomic reference and used to manually segment gray and white matter in the brain sections.

**Noise Experiments**

A total of six healthy subjects (age 24–34 years, 28 ± 4 years) participated in the noise imaging procedure at the 3.0 T scanner. The human protocol was approved by the Institutional Review Board of the Stanford University School of Medicine. Informed consent was obtained prior to the examinations. The subjects were instructed to close their eyes and relax in the magnet. In each session, five 7.5-min protocols with the dual-echo sequence were performed resulting in a total of 1200 frames (150 frames per slice and echo) from each trial. In order to separate the signal-dependent physiological noise \( \sigma_r \) from the intrinsic raw noise \( \sigma_0 \), we modulated the MR signal strength dynamically by alternating the flip angles between \( 19^\circ, 42^\circ, \) and \( 90^\circ \) in an interleaved fashion. These particular flip angles were picked to obtain \( \frac{1}{2} \lambda \), and the maximum signal. To investigate simultaneously the echo-time dependency of the physiological noise, TE combinations of 9.0 and 30.0 ms, 30.0 and 52.0 ms, 52.0 and 74.0 ms, 74.0 and 96.0 ms, as well as 9.0 and 96.0 ms were performed. Thus, each protocol with a total of 1200 frames comprised the acquisition of 50 frames per slice, echo, and flip angle.

**Signal-to-Noise and Physiological Noise**

The SNR and noise \( \alpha \) of \( T_2^* \)-weighted images were determined by calculating pixel-wise the mean signal intensity and standard deviation in the time series of 50 respective frames. The raw noise \( \sigma_0 \) (signal-independent noise) was determined by extrapolating from the total noise at three different flip angles to the noise at a flip angle of \( \alpha = 0^\circ \). The slope of the interpolated graph determines \( \lambda \) in Eq. [4]. The noise contributions \( \sigma_p \) and \( \sigma_{NB} \) in Eq. [3] were determined from the calculated \( \lambda \)-s at five different echo times, i.e., intrinsic \( T_2^* \) values were calculated pixelwise by means of a least-squares fit of Eq. [2] to the experimental values. The SNR as well as the different noise contributions \( \sigma, \sigma_0, \sigma_p, \) and \( \sigma_{NB} \) were determined in manually drawn gray and white matter regions-of-interest (ROIs).

**Phantom Noise**

Identical experiments were also performed on a 17 cm diameter spherical water phantom doped with nickel-chloride in order to further characterize and validate the different noise contributions. We expected that \( \sigma_p \) is negligible in the phantom.
Table 1  
Gray Matter Region

<table>
<thead>
<tr>
<th>Subject</th>
<th>SNR</th>
<th>$S_0$</th>
<th>$\sigma$</th>
<th>$\sigma_a$</th>
<th>$\sigma_B$</th>
<th>$\sigma_{NB}$</th>
<th>$\lambda$ [10^{-5}]</th>
<th>$T_2^* [\text{ms}]$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>101.4</td>
<td>3261</td>
<td>0.54</td>
<td>0.21</td>
<td>0.42</td>
<td>0.27</td>
<td>0.9</td>
<td>51.8</td>
</tr>
<tr>
<td>2</td>
<td>57.9</td>
<td>4429</td>
<td>0.86</td>
<td>0.19</td>
<td>0.66</td>
<td>0.51</td>
<td>1.7</td>
<td>43.3</td>
</tr>
<tr>
<td>3</td>
<td>67.4</td>
<td>3577</td>
<td>0.74</td>
<td>0.24</td>
<td>0.69</td>
<td>0.16</td>
<td>1.4</td>
<td>48.4</td>
</tr>
<tr>
<td>4</td>
<td>96.4</td>
<td>2802</td>
<td>0.53</td>
<td>0.22</td>
<td>0.37</td>
<td>0.31</td>
<td>1.0</td>
<td>48.8</td>
</tr>
<tr>
<td>5</td>
<td>98.1</td>
<td>4995</td>
<td>0.55</td>
<td>0.17</td>
<td>0.52</td>
<td>0.0</td>
<td>1.0</td>
<td>45.0</td>
</tr>
<tr>
<td>6</td>
<td>82.9</td>
<td>2768</td>
<td>0.54</td>
<td>0.21</td>
<td>0.50</td>
<td>0.0</td>
<td>1.1</td>
<td>41.0</td>
</tr>
<tr>
<td>Mean</td>
<td>84 ± 18</td>
<td>3538 ± 740</td>
<td>0.63 ± 0.14</td>
<td>0.21 ± 0.02</td>
<td>0.53 ± 0.13</td>
<td>0.21 ± 0.20</td>
<td>1.2 ± 0.3</td>
<td>46.3 ± 3.9</td>
</tr>
<tr>
<td>Phantom</td>
<td>338</td>
<td>1137</td>
<td>0.19</td>
<td>0.14</td>
<td>0.08</td>
<td>0.08</td>
<td>0.2</td>
<td>72.8</td>
</tr>
</tbody>
</table>

The SNR, the signal $S_0$, the intrinsic noise $\sigma$, the raw noise $\sigma_0$, the physiological noise contributions $\sigma_B$ and $\sigma_{NB}$, the $\lambda$-constant, and $T_2^*$-values in gray matter regions from all six subjects after the retrospective correction of physiological motion effects. The last row in Table 1 shows the respective values as determined in phantom experiments. $S_0$ demonstrates the extrapolated signal strengths at TE = 0 ms. The SNR, noise values, and $\lambda$ represent the respective values at TE = 30 ms. Note that $\sigma$ reflects the standard deviation in the time series and $\sigma_0$, $\sigma_B$, and $\sigma_{NB}$ are determined using the model in Eqs. [1]--[4], i.e., SNR and the signal-dependent noise values $\sigma_0$ and $\sigma_{NB}$ reflect results with $\alpha = 90^\circ$. For intersubject comparison noise values were normalized by the signal intensity $S_0$ and expressed as percentage of $S_0$.

**RESULTS**

**Signal-to-Noise and Image Noise**

The SNR, the signal intensity $S$ at TE = 0 ms, the intrinsic noise $\sigma$, the raw noise $\sigma_0$, the physiological noise contributions $\sigma_B$ and $\sigma_{NB}$, and $T_2^*$-values in gray matter regions from all six subjects after the retrospective correction of physiological motion effects. The last row in Table 1 shows the respective values as determined in phantom experiments. $S_0$ demonstrates the extrapolated signal strengths at TE = 0 ms. The SNR, noise values, and $\lambda$ represent the respective values at TE = 30 ms. Note that $\sigma$ reflects the standard deviation in the time series and $\sigma_0$, $\sigma_B$, and $\sigma_{NB}$ are determined using the model in Eqs. [1]--[4], i.e., SNR and the signal-dependent noise values $\sigma_0$ and $\sigma_{NB}$ reflect results with $\alpha = 90^\circ$. For intersubject comparison noise values were normalized by the signal intensity $S_0$ and expressed as percentage of $S_0$.

A spatial analysis of the noise contributions after the retrospective correction demonstrates that the raw noise $\sigma_0$ is almost identical in gray and white matter. In contrast to $\sigma_0$, the total image noise $\sigma$ is found to be 1.7× greater in gray matter than in white matter regions and accounts for the observation of a significantly ($P \leq 0.01$) smaller average SNR in gray matter (SNR = 84 ± 18) than in white matter regions (SNR = 143 ± 30). Similarly, the physiological noise contribution $\sigma_B$ is on average 2.1× higher in gray matter than in white matter regions and in gray matter consistently stronger than $\sigma_0$. The physiological noise contribution $\sigma_{NB}$ is about 1.9× greater in gray matter regions than in respective white matter, but in magnitude a factor $>2$ smaller than $\sigma_B$.

Figure 1a demonstrates the physiological noise contribution $\sigma_B$ in cortical gray matter regions as a function of TE averaged over all subjects. The solid line represents the best fit of Eq. [2] to the average $\sigma_B$ ($n = 6$). The results are normalized with respect to the maximum $\sigma_B$ as determined by the fit. Figure 1b shows the $\sigma_B$'s from all six subjects and the best fit to the respective values. In order to emphasize the strong influence of the physiological noise to the total image noise, the individual values are normal-

Table 2  
White Matter Region

<table>
<thead>
<tr>
<th>Subject</th>
<th>SNR</th>
<th>$S_0$</th>
<th>$\sigma$</th>
<th>$\sigma_a$</th>
<th>$\sigma_B$</th>
<th>$\sigma_{NB}$</th>
<th>$\lambda$ [10^{-5}]</th>
<th>$T_2^* [\text{ms}]$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>165.0</td>
<td>3152</td>
<td>0.31</td>
<td>0.21</td>
<td>0.18</td>
<td>0.14</td>
<td>0.5</td>
<td>51.6</td>
</tr>
<tr>
<td>2</td>
<td>104.5</td>
<td>4482</td>
<td>0.48</td>
<td>0.19</td>
<td>0.33</td>
<td>0.28</td>
<td>0.9</td>
<td>44.0</td>
</tr>
<tr>
<td>3</td>
<td>107.4</td>
<td>3251</td>
<td>0.48</td>
<td>0.26</td>
<td>0.40</td>
<td>0.0</td>
<td>0.8</td>
<td>49.4</td>
</tr>
<tr>
<td>4</td>
<td>160.2</td>
<td>2604</td>
<td>0.33</td>
<td>0.23</td>
<td>0.17</td>
<td>0.15</td>
<td>0.5</td>
<td>50.9</td>
</tr>
<tr>
<td>5</td>
<td>174.0</td>
<td>4154</td>
<td>0.32</td>
<td>0.20</td>
<td>0.24</td>
<td>0.0</td>
<td>0.5</td>
<td>51.4</td>
</tr>
<tr>
<td>6</td>
<td>150.5</td>
<td>2466</td>
<td>0.32</td>
<td>0.24</td>
<td>0.18</td>
<td>0.11</td>
<td>0.5</td>
<td>46.5</td>
</tr>
<tr>
<td>Mean</td>
<td>143 ± 30</td>
<td>3351 ± 814</td>
<td>0.37 ± 0.08</td>
<td>0.22 ± 0.03</td>
<td>0.25 ± 0.10</td>
<td>0.11 ± 0.11</td>
<td>0.6 ± 0.2</td>
<td>49.0 ± 3.1</td>
</tr>
</tbody>
</table>

The SNR, the signal $S_0$, the intrinsic noise $\sigma$, the raw noise $\sigma_0$, the physiological noise contributions $\sigma_B$ and $\sigma_{NB}$, the $\lambda$-constant, and $T_2^*$-values in white matter regions from all six subjects after the retrospective correction of physiological motion effects. The SNR, noise values, and $\lambda$ represent the respective values at TE = 30 ms. For intersubject comparison noise values were normalized by the signal intensity $S_0$ and expressed as percentage of $S_0$. 

Physiological Noise in MRI
Note that results from an individual subject represent an average value over four slices and two trials per echo time. The corresponding results in white matter regions are shown in Fig. 2a,b.

Spatial Distribution of Noise Components
The spatial distributions of several noise sources in an exemplary slice from one subject and in the phantom are shown in Fig. 3. Figure 3a demonstrates a high resolution MRI of the section. Figure 3b–d represents the raw noise $σ_0$, the physiological noise contribution $σ_B$, as well as the noise contribution $σ_{NB}$ in the same slice at TE = 30 ms. Both the $σ_0$-map (Fig. 3b) and the $σ_{NB}$-map (Fig. 3d) demonstrate a rather uniform distribution of noise. However, the $σ_{NB}$-map (Fig. 3c) clearly exhibits spatial differences in the noise, showing larger noise in gray matter regions. Higher noise values at the frontal brain area in the $σ_{NB}$-map (Fig. 3d) presumably reflect slight subject motion. The phantom raw noise in Fig. 3e exhibits a slight spatial gradient that may reflect a $B_1$ heterogeneity from the RF-coil sensitivity profile. The focal regions in the respective

FIG. 1. The mean ($n = 6$) physiological noise contribution $σ_B$ as a function of TE in cortical gray matter regions (a). For the intersubject comparison, $σ_B$'s were individually normalized by the interpolated noise at TE = $T_2\ast$ prior to averaging. The symbols represent the mean noise and the solid line shows the best fit of Eq. [2]. Errors bars (±SD) represent the variability across subjects. b: $σ_B$ as a function of TE from all six subjects. Here, the individual graphs were normalized by the respective raw noise $σ_0$. 

FIG. 2. The corresponding physiological noise contribution $σ_B$ in white matter.
The $\sigma_B$-map (Fig. 3f) are presumably due to vibrations of water in the phantom near regions of altered susceptibility from a label attached to the phantom and a plug that fills a hole used for filling the phantom with water. However, in this work we propose only a simple model of the noise in GRE imaging, and therefore such effects, as well as obvious vascularity (e.g., in the center of the anatomical slice in Fig. 3a), violates this simple model.

**DISCUSSION**

The proposed model in Eqs. [1]–[4] for SNR and noise properties in gradient-echo (GRE) MRI suggests that the physiological noise increases with the signal strength and limits the maximum attainable SNR at a ratio of $1/\lambda$. The mean $\lambda$ values across six subjects, e.g., at TE $= 30$ ms, of $\lambda_{gm} = 0.012 \pm 0.003$ and $\lambda_{wm} = 0.006 \pm 0.002$ predict that the maximum attainable SNR in gray matter regions amounts to SNR $= 83$ in cortical gray matter and to SNR $= 167$ in white matter regions for the acquisition parameters and hardware used here. The respective SNR of 84 and 143 obtained in the present study are close to these predicted asymptotic SNR-values of $\sim 1/\lambda$ and imply that further improvements in coil and magnetic field strength would yield only moderate additional gains in SNR. It can be argued that TR $= 3.0$ sec is not sufficient to suppress the
development of steady-state magnetization in the post-90°-pulse images in voxels that are contaminated by CSF. However, in a recent study (8) we investigated the physiological noise term \( \sigma_p \) with repetition times of TR = 3.0 ms and TR = 5.4 sec. The finding of very similar \( \lambda \)'s at both TRs demonstrates that the calculated noise terms can be considered to be nearly uninfluenced by saturation effects. This result is because the CSF signal was greatly suppressed by the use of RF spoiling, and contamination from CSF was minimized.

The retrospective correction scheme applied here (18), corrects effectively for respiratory- and cardiac-induced periodic fluctuations in \( R_2^* \) and reduces the corresponding noise energy to the average noise level. More explicitly, we observed a 7% decrease in \( \sigma_0 \) and a 10% decrease in \( \sigma_B \). However, fluctuations in brain metabolism and neuronal activity and the related physiological changes in CBF, CBV, and CMRO2 cause signal changes due to a true BOLD-effect and are unaffected by those correction schemes. Considering the favorable motion insensitivity of the BOLD-effect and are unaffected by the two noise sources \( \sigma_B \) and \( \sigma_{NB} \), as the TE-dependent physiological noise contribution theoretically disappears in spin-echo images.

Recently published preliminary results from a comparison of SNR and noise properties at 1.5 T and 3.0 T using conventional EPI (22) have also reported that physiological noise appears to be more dominant at higher fields. In that study, the signal strength was modulated by a variation of TE. The present results, however, clearly suggest that the physiological noise consists of at least two different contributions, one of them with a TE-dependency. Therefore, it is necessary to modulate the signal without changing TE (e.g., using the flip angle as in this work) in addition to obtaining measurements with different TE’s to separate the two noise components.

**CONCLUSIONS**

The present results demonstrate that the image noise in oxygenation-sensitive neuroimaging at 3.0 T is decisively dominated by the physiological noise. This is due to the fact that \( \sigma_B \), \( \sigma_{NB} \), and \( \sigma_{np} \) are signal-dependent and increase with improved RF-coils and field strength. \( \sigma_p \), which demonstrates the primary contributor to \( \sigma_B \), is found to be 2.1× higher in cortical grey matter than in white matter regions and therefore limits the maximum attainable SNR and CNR in MRI at higher fields. However, this noise source also may reveal important information on brain physiology, as it represents fluctuations in CBF, CBV, and oxidative metabolism. Strategies such as investigations of functional connectivities and event-related fMRI, which analyze spatial and temporal correlations of signal fluctuation time courses (noise), are predicted to benefit from the increased influence of \( \sigma_p \) at higher fields.

**ACKNOWLEDGMENTS**

The authors thank A.M. Sawyer-Glover for assistance and T.J. Brosnan for technical support.

**REFERENCES**