

and

$$\vec{m}_l(t) \equiv \int_0^t \gamma \vec{G}(\tau) \frac{\tau^l}{l!} d\tau \quad l = 0, 1, 2, \dots \quad (4.89)$$

$\vec{m}_l$ is the l^{th} order gradient moment.

Proof of Eq. 4.87. The exact path $\vec{r}(t)$ followed by the moving spin is unknown. However, any physical motion can be expanded in a Taylor series around $t = 0$. Hence,

$$\vec{r}(t) = \vec{r}(0) + \frac{d\vec{r}}{dt}(0)t + \dots + \frac{d^l \vec{r}}{dt^l}(0) \frac{t^l}{l!} + \dots \quad (4.90)$$

The position $\vec{r}$, the velocity $\vec{v}(\vec{r})$, and the acceleration $\vec{a}(\vec{r})$ of the spin at time $t = 0$ can be introduced in this equation:

$$\vec{r}(t) = \vec{r} + \vec{v}(\vec{r}) t + \frac{\vec{a}(\vec{r}) t^2}{2} + \dots \quad (4.91)$$

Substituting Eq. 4.91 in Eq. 4.86 yields:

$$\begin{aligned} \Phi(\vec{r}, t) = & \vec{r} \cdot \int_0^t \gamma \vec{G}(\tau) d\tau + \vec{v}(\vec{r}) \cdot \int_0^t \gamma \vec{G}(\tau) \tau d\tau \\ & + \vec{a}(\vec{r}) \cdot \int_0^t \gamma \vec{G}(\tau) \frac{\tau^2}{2} d\tau + \dots \end{aligned}$$

or, using the gradient moments as defined in Eq. 4.89

$$\Phi(\vec{r}, t) = \vec{r} \cdot \vec{m}_0(t) + \vec{v}(\vec{r}) \cdot \vec{m}_1(t) + \vec{a}(\vec{r}) \cdot \vec{m}_2(t) + \dots \quad (4.92)$$

Rewriting Eq. 4.47 as

$$s(t) = \int_{\vec{r}} \rho^*(\vec{r}) e^{-i\Phi(\vec{r}, t)} d\vec{r}, \quad (4.93)$$

and substituting Eq. 4.92 into Eq. 4.93, yields Eq. 4.87.

Without motion, only the zeroth-order moment $\vec{m}_0(t)$ in Eq. 4.87 causes a phase shift. This phase shift is needed for position encoding when using the $\vec{k}$ -theorem. Motion introduces additional dephasing of the signal $s(t)$, yielding distortion and contrast loss. However, motion-induced dephasing can be reduced by *back-to-back symmetric bipolar pulses of opposite polarity*. They are able to restore hyperintense vessel signals for blood flowing at a *constant* velocity. In case of constant velocity Eq. 4.87 becomes

$$s(t) = \int_{\vec{r}} \rho^*(\vec{r}) e^{-i\vec{v}(\vec{r}) \cdot \vec{m}_1(t)} e^{-i\vec{r} \cdot \vec{m}_0(t)} d\vec{r} \quad (4.94)$$

and contains only two dephasing factors, one necessary for position encoding and the other introduced by the blood velocity $\vec{v}(\vec{r})$.

Eq. 4.55 shows that for *stationary* spins ($\vec{v}(\vec{r}) = 0$) the net phase shift due to simple bipolar gradient pulses (Figure 4.27(a)) is zero. This is the case at $t = \text{TE}$ in the frequency-encoding and slice-selection directions. For *moving* spins ($\vec{v}(\vec{r}) \neq 0$), however, a simple bipolar pulse sequence as in Figure 4.27(a) introduces a phase shift because its first gradient moment $\vec{m}_1$ at $t = \text{TE}$ is nonzero:

$$m_1(\text{TE}) = -\gamma \vec{G}(\Delta t)^2 \neq 0. \quad (4.95)$$

Back-to-back symmetric bipolar pulses of opposite polarity on the other hand (Figure 4.27(b)) remove the velocity-induced phase shift at $t = \text{TE}$ while they have no net effect on static spins. Both their zeroth and first-order gradient moments $m_0(\text{TE})$ and $m_1(\text{TE})$ are zero. Higher order motion components are *not* rephased, however, and will still cause dephasing.

The rephasing gradients can be applied only in the frequency-encoding and slice-selection directions. This technique is known as *gradient moment nulling*, *gradient moment rephasing*, or *flow compensation*. It is built-in in sequences to remove velocity-induced artifacts. A diagram of a 3D spoiled GE sequence with first-order motion compensation is shown in Figure 4.28. Technical considerations limit the motion compensation to the first-order or at most the second-order gradient moments. Very complex motion patterns, such as the turbulence in the aortic arch and behind stenotic plaques, continue to produce signal dephasing.

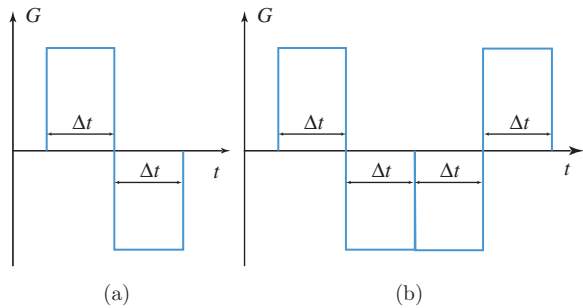


Figure 4.27 (a) Simple bipolar pulses cannot provide a phase-coherent signal for moving spins. (b) Back-to-back symmetric bipolar pulses of opposite polarity on the other hand restore the phase coherence completely for spins moving at a constant velocity.