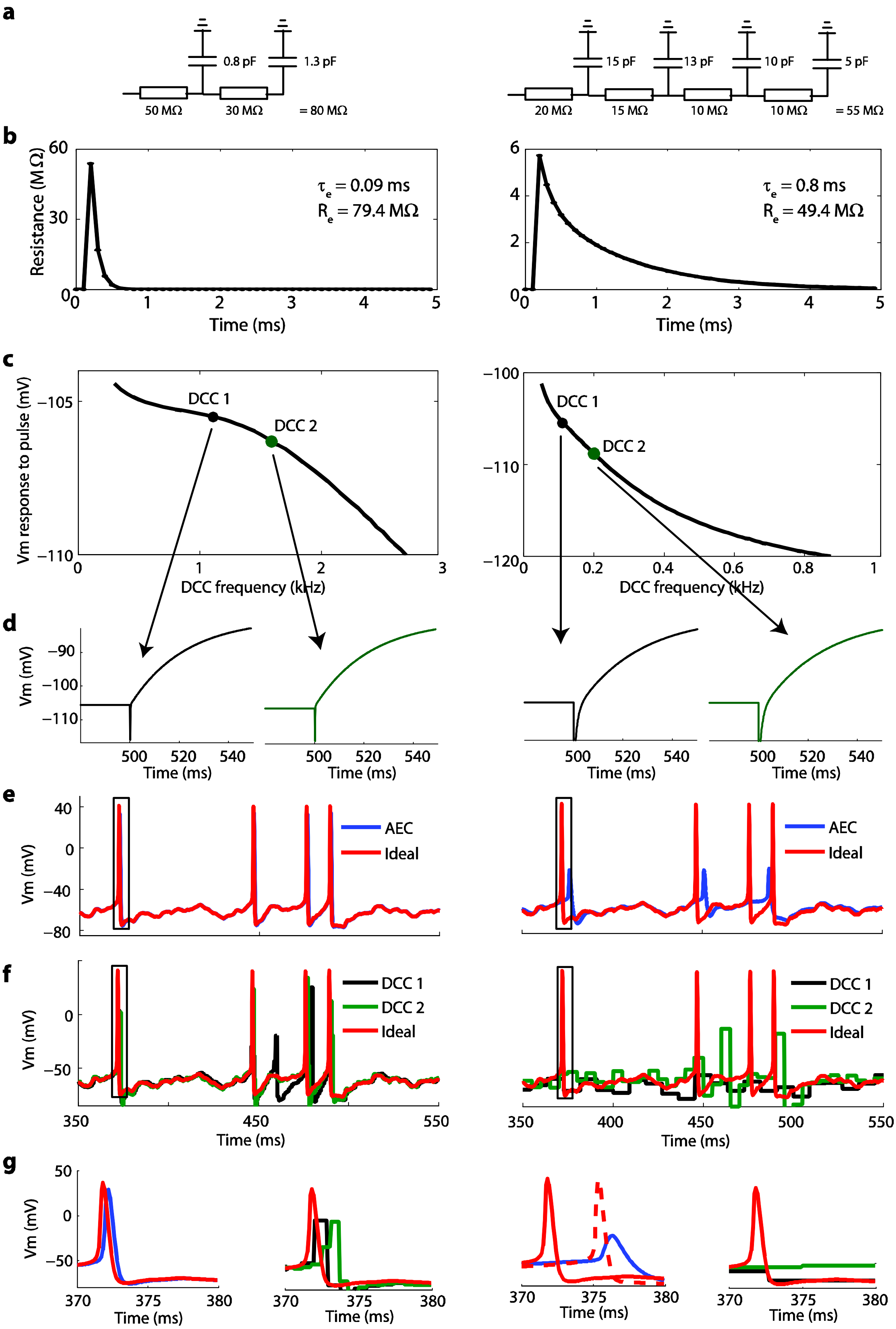
Supplementary Figure 3 – Numerical simulation of conductance noise injection with AEC and DCC.



Left column : with a fast electrode made of two resistors and capacitors (time constant 0.09 ms from exponential fit); right column: with a slow electrode made of four resistors and capacitors (time constant 0.8 ms). The electrode is impaled into a cortical cell modeled as a single-compartment Hodgkin-Huxley type model (equations and parameters in Destexhe et al, J Neurophysiol 79 (1998), 999-1016), and connected to a model amplifier (emulating DCC, bridge compensation and an acquisition board; we used a model of the electrode previously published in a conference proceedings - Brette et al, Neurocomputing 70 (2007), 1597-1601 proceedings of CNS 2006, Edinburgh, UK). Programs were written in C++, differential equations were integrated with a second-order Runge-Kutta method (time step 1 µs).

a) Electrical circuit of the electrode. The time constant of the fast electrode corresponds to our measurements in vitro (with full capacitance neutralization).

b) Electrode kernel estimated in the cell with AEC, and estimation of the electrode resistance (0.75% error for the fast electrode, 10% error for the slow electrode).

c) Vm response in DCC mode to a -0.4 nA current pulse lasting 400 ms as a function of the DCC frequency. DCC 1 is the optimal setting (left: 1.1 kHz, right: 0.11 kHz) for which the recorded response corresponds to the true intracellular Vm. The optimal DCC period is typically 8-10 times the electrode time constant. DCC 2 is a slightly higher setting, which corresponds to an underestimation of the electrode resistance (left: 1.6 kHz, Re = 78 MΩ; right: 0.2 kHz, Re = 47.5 MΩ). With the fast electrode, the inflexion point is close to the optimal setting but is not identical; with the slow electrode, there is no inflexion point.

d) Bridge balancing traces for the corresponding settings (Vm recording of a pulse response using bridge compensation, i.e., estimating the electrode response as $R_e \times I(t)$). Black = optimal setting (perfect bridge compensation), green = higher setting, i.e., underestimation of the resistance. With the fast electrode, the non-optimal setting is hard to distinguish from the optimal one; with the slow electrode, it is not possible.

e) Response to a conductance noise injection (see main text) using AEC in dynamic-clamp. The red trace is the ideal recording with a zero-resistance electrode; the blue trace is the recording with AEC. The spike in the rectangle is enlarged in (g). With the fast electrode, both the subthreshold response and spikes are obtained with an excellent quality. With the slow electrode, the subthreshold response is good and spikes are recorded, but in a filtered version (see (g)).

f) Response to the same injection with DCC in dynamic-clamp. With the fast electrode, subthreshold responses are approximately correct (but a low temporal resolution and high-frequency noise) but spikes are missed or distorted. Besides, the setting which was optimal for current pulses (DCC 1) does not seem optimal for nonlinear events. With the slow electrode, both recordings are meaningless. Increasing the DCC frequency leads to unstable oscillations like in bridge (not shown).

g) Zoom on spikes (enlargement of rectangles in (e) and (f)). With the fast electrode, spikes recorded with AEC are very well reproduced with a slight filtering, while spikes recorded with DCC are very distorted due to the low temporal resolution. With the slow electrode, spikes recorded with AEC are filtered and delayed. The delay is also present in the cell (dashed red line = real Vm during AEC recording) and is due to a higher injected current (because the electrode resistance is underestimated); it does not appear in current clamp mode (not shown). Spikes cannot be recorded with DCC. Thus AEC recordings are possible with a slow electrode (equivalently, with a fast membrane) where DCC fails.